LASER MICROMACHINING OF CORONARY STENTS FOR MEDICAL APPLICATIONS

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# Table of Contents

Table of Contents ....................................................................................................... 2  
Abstract ....................................................................................................................... 6  
List of Figures ............................................................................................................. 7  
List of Tables ............................................................................................................ 14  
Declaration ................................................................................................................ 15  
Copyright Statement ................................................................................................ 16  
Acknowledgements ................................................................................................... 17  
List of Publications and Award .............................................................................. 18  
Nomenclature ............................................................................................................ 20

## Chapter 1

**Introduction**.............................................................................................................. 23  
1.1 Research motivation and rationale ............................................................. 23  
1.2 Aim and objectives ..................................................................................... 25  
1.2.1 Aim ......................................................................................................... 25  
1.2.2 Objectives ............................................................................................... 25  
1.3 Thesis outline ............................................................................................. 26

## Chapter 2

**Review of Laser Cutting Fundamentals**................................................................ 29  
2.1 Introduction ................................................................................................ 29  
2.2 Laser cutting introduction .......................................................................... 29  
2.2.1 The advantages of laser cutting .............................................................. 31  
2.2.2 Laser cutting mechanisms ...................................................................... 32  
2.2.3 Laser cutting process parameters ........................................................... 33  
2.2.4 Characteristics of laser cut materials ...................................................... 38  
2.3 Types of lasers ............................................................................................ 42  
2.3.1 CO₂ laser ................................................................................................ 43  
2.3.2 Nd:YAG laser .......................................................................................... 43  
2.3.3 Fibre laser ............................................................................................... 44  
2.3.4 Ultra short pulse lasers ........................................................................... 47  
2.4 Summary .................................................................................................... 48

## Chapter 3

**Laser cutting of coronary stents for cardiovascular treatments – process characteristics and challenges**................................................................................. 49  
3.1 Introduction ................................................................................................ 49  
3.2 Coronary stents ........................................................................................... 50  
3.3 Bare Metal Stents (BMS) and Drug Eluting Stents (DES) ................................ 52  
3.4 Stent Materials ............................................................................................ 53  
3.5 Stent fabrication methods ........................................................................... 55  
3.6 Laser Cutting of Coronary Stents .................................................................. 58
Table of Contents

3.6.1 Laser stent cutting – basic process description ............................................. 58
3.6.2 Important quality aspects in laser stent cutting ........................................... 60
3.6.3 Process parameters affecting laser stent cutting ...................................... 61
3.7 Progress and Development in Laser Based Stent Cutting Technology........... 67
3.8 Approaches in improving the laser stent cutting process ......................... 73
3.8.1 Water assisted laser cutting of stents ......................................................... 73
3.8.2 Other methods ....................................................................................... 73
3.9 Stent finishing: post-processing stages in laser stent manufacturing .......... 74
3.10 Challenges and Future Prospects ............................................................... 77
3.11 Summary .................................................................................................... 80

Chapter 4
Experimental Apparatus and Analytical Equipment ......................................... 81
4.1 Introduction .................................................................................................. 81
4.2 Stent cutting system – Micro T15 ............................................................... 82
4.2.1 Fibre laser .............................................................................................. 83
4.2.2 Picosecond laser .................................................................................... 83
4.3 Femtosecond laser system .......................................................................... 84
4.4 Analytical equipment ................................................................................ 86
4.4.1 Optical microscope ................................................................................ 86
4.4.2 Scanning electron microscope (SEM) ................................................... 87
4.4.3 White light optical interferometer ......................................................... 87
4.4.4 MTS Nano Indenter XP ....................................................................... 89
4.5 Summary ..................................................................................................... 90

Chapter 5
Comparison of dry and wet fibre laser profile cutting of thin 316L stainless steel tubes for medical device applications ......................................................... 91
5.1 Introduction ................................................................................................ 92
5.2 Experimental procedures .......................................................................... 93
5.2.1 The stent cutting system ....................................................................... 93
5.2.2 Materials ............................................................................................... 94
5.2.3 Cutting experiments ............................................................................. 94
5.3 Results ........................................................................................................ 97
5.3.1 Effects of cutting parameters upon kerf width and surface roughness .. 97
5.3.2 Dross formation ................................................................................... 99
5.3.3 Back wall damage ................................................................................. 100
5.3.4 Heat effects ......................................................................................... 101
5.4 Discussions ............................................................................................... 103
5.5 Conclusions ............................................................................................. 108

Chapter 6
Picosecond laser micromachining of nitinol and platinum iridium alloy for coronary stent applications ................................................................. 109
6.1 Introduction ............................................................................................... 110
6.2 Experimental Procedures ........................................................................ 111
Table of Contents

6.2.1 Materials............................................................................................... 111
6.2.2 Equipment ............................................................................................ 111
6.2.3 Laser Cutting ........................................................................................ 112
6.3 Results ...................................................................................................... 114
  6.3.1 SEM images of the laser cut samples ................................................... 114
  6.3.2 Demonstration of dross free cut ........................................................... 114
  6.3.3 Topography analysis of the cut surface ................................................ 116
  6.3.4 Striations formation on the cut surface ................................................ 117
  6.3.5 HAZ analysis ........................................................................................ 118
6.4 Discussions ............................................................................................... 121
6.5 Conclusions .............................................................................................. 126

Chapter 7
Underwater femtosecond laser micromachining of thin nitinol tubes for medical coronary stent manufacture .................................................................................. 128
7.1 Introduction .............................................................................................. 129
7.2 Experimental procedures .......................................................................... 131
  7.2.1 Experimental setup ............................................................................... 131
7.3 Results ...................................................................................................... 136
  7.3.1 Number of passes to achieve full-depth penetration cut ......... 136
  7.3.2 The effect of the process parameters on the cut kerf width .............. 137
  7.3.3 The effect of the process parameters on the cut surface quality ........ 138
  7.3.4 Heat Affected Zone (HAZ) analysis ..................................................... 143
  7.3.5 Debris and recast formation ................................................................. 145
7.4 Discussions ............................................................................................... 147
  7.4.1 Debris and recast formation mechanism in dry cutting ............... 147
  7.4.2 Approach in preventing recast and debris formation ................. 151
7.5 Conclusions .............................................................................................. 153

Chapter 8
Understanding the behaviour of pulsed laser dry and wet micromachining processes by multi-phase smoothed particle hydrodynamics (SPH) modelling 154
8.1 Introduction .............................................................................................. 155
8.2 Model description ..................................................................................... 157
  8.2.1 Physical process to be captured in SPH model ......................... 157
  8.2.2 Experimental procedures for verification ........................................ 159
  8.2.3 Physical Properties ......................................................................... 160
8.3 Mathematical formulation ........................................................................ 161
  8.3.1 Model assumptions ........................................................................ 161
  8.3.2 Heat transfer .................................................................................... 161
  8.3.3 Melt ejection velocity ................................................................. 162
  8.3.4 Vapour pressure .............................................................................. 162
  8.3.5 Effective assist gas pressure .......................................................... 163
  8.3.6 Modelling the phase changes ......................................................... 163
8.4 SPH Model – SPHysics ............................................................................ 164
  8.4.1 Governing equations ...................................................................... 164
<table>
<thead>
<tr>
<th>Section</th>
<th>Title</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>8.4.2</td>
<td>Boundary conditions</td>
<td>167</td>
</tr>
<tr>
<td>8.4.3</td>
<td>Detection of surface particle</td>
<td>168</td>
</tr>
<tr>
<td>8.4.4</td>
<td>SPH simulation procedure</td>
<td>168</td>
</tr>
<tr>
<td>8.5</td>
<td>Results and Discussions</td>
<td>170</td>
</tr>
<tr>
<td>8.5.1</td>
<td>Penetration depth</td>
<td>170</td>
</tr>
<tr>
<td>8.5.2</td>
<td>Phase changes</td>
<td>173</td>
</tr>
<tr>
<td>8.5.3</td>
<td>Melt ejection velocity</td>
<td>174</td>
</tr>
<tr>
<td>8.5.4</td>
<td>Formation of recast and spatter</td>
<td>175</td>
</tr>
<tr>
<td>8.5.5</td>
<td>Backwall damage and effect of water flow in thin tube cutting</td>
<td>177</td>
</tr>
<tr>
<td>8.6</td>
<td>Conclusions</td>
<td>180</td>
</tr>
</tbody>
</table>

**Chapter 9**

**Conclusions and Future Work**

<table>
<thead>
<tr>
<th>Section</th>
<th>Title</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>9.1</td>
<td>Conclusions</td>
<td>182</td>
</tr>
<tr>
<td>9.1.1</td>
<td>Experimental work</td>
<td>183</td>
</tr>
<tr>
<td>9.1.2</td>
<td>SPH modelling</td>
<td>185</td>
</tr>
<tr>
<td>9.2</td>
<td>Recommendations for future work</td>
<td>186</td>
</tr>
<tr>
<td>9.2.1</td>
<td>Experimental work</td>
<td>186</td>
</tr>
<tr>
<td>9.2.2</td>
<td>SPH modelling</td>
<td>187</td>
</tr>
</tbody>
</table>

**References**

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Abstract

This PhD thesis reports an investigation into medical coronary stent cutting using three different types of lasers and associated physical phenomena. This study is motivated by a gap in the current knowledge in stent cutting identified in an extensive literature review. Although lasers are widely used for stent cutting, in general the laser technology employed is still traditionally based on millisecond pulsed Nd:YAG lasers. Although recent studies have demonstrated the use of fibre lasers, picosecond and femtosecond lasers for stent cutting, it has been preliminary studies.

To further understand the role of new types of lasers such as pulsed fibre lasers, picosecond and femtosecond pulsed lasers in stent cutting, these three lasers based stent cutting were investigated in this project. The first investigation was on a new cutting method using water assisted pulsed (millisecond) fibre laser cutting of stainless steel 316L tubes to explore the advantages of the presence of water compared to the dry cutting condition. Significant improvements were observed with the presence of water; narrower kerf width, lower surface roughness, less dross attachment, absence of backwall damage and smaller heat affected zone (HAZ). This technique is now fully commercialised by Swisstec, an industrial project partner that manufactures stent cutting machines.

The second investigation used the picosecond laser (with 6 ps pulse duration in the UV wavelength range) for cutting nickel titanium alloy (nitinol) and platinum iridium alloy. The main achievement in this study was obtaining dross-free cut as well as clean backwall, which may eliminate the need for extensive post-processing. Picosecond laser cutting of stents is investigated and reported for the first time.

The third area of investigation was on the use of a femtosecond laser at 100 fs pulse duration for cutting nickel titanium alloy tubes. It was found that dry cutting degraded the cut quality due to debris and recast formation. For improvement, a water assisted cutting technique was undertaken, for the first time, by submerging the workpiece in a thin layer of water for comparison with the dry cutting condition.

The final part of the thesis presents a three dimensional numerical model of the laser micromachining process using smoothed particle hydrodynamics (SPH). The model was used to provide better understanding of the laser beam and material interaction (with static beam) including the penetration depth achieved, phase changes, melt ejection velocity, also recast and spatter formation. Importantly, the model also simulated the wet machining condition by understanding the role of water removing the melt ejected during the process which avoided backwall damages. Results with the fibre laser in millisecond pulse duration were used for the validation purposes. The conclusions reached in this project and recommendations for future work are enclosed.

The work has resulted in the publication of 3 journal papers and 2 additional journal paper submissions.
List of Figures

Figure 2.1: Schematic diagram of the laser cutting process ................................. 30
Figure 2.2: Laser cutting approaches: (a) fusion cutting, (b) reactive fusion cutting and (c) vaporisation cutting. .............................................................. 32
Figure 2.3: The acceptable-quality cutting region of 3 mm mild steel oxygen-assisted laser cutting ................................................................. 37
Figure 2.4: Dross formation .................................................................................... 39
Figure 2.5: (a) SEM photograph and hardness variation of the transverse section of a typical laser cut surface ...................................................... 40
Figure 2.6: Mean height of the profile ................................................................. 41
Figure 2.7: Striations appear in the form of periodic lines ................................. 42
Figure 2.8: A schematic drawing of the double-clad fibre .................................. 45
Figure 2.9: Configuration of end pumped fibre laser ........................................ 45
Figure 2.10: Configuration of multi-mode pumped fibre laser ......................... 46
Figure 2.11: The chirped pulse amplification (CPA) technique for ultra-short pulse laser. The given pulse durations are typical values for presently applied femtosecond laser systems ......................................................... 48
Figure 3.1: An illustration of blood vessel blockage and coronary stent placement. (a) An artery (with plague build up) inserted with a cramped stent mounted on a balloon catheter (b) The balloon is inflated and widening the artery and (c) Widened-artery restored normal blood flow (Photo from National Heart Lung and Blood Institute) ................................................. 51
Figure 3.2: Schematic of Bare Metal Stent and TAXUS stent (Drug-eluting stent by Boston Scientific) ............................................................................. 52
Figure 3.3: Example of stents fabricated with traditional methods. (a) Braided stent fabricated from cobalt alloy wire, (b) Knitted stent made of tantalum wire and (c) Welded stent made of nitinol wire ........................................... 55
Figure 3.4: Example of coronary stent cut by laser ............................................. 56
Figure 3.5: Optimised stent design (BIOTRONIK, Tenax 15) ........................................58
Figure 3.6: Schematically drawn setup of a laser-based stent cutter .........................59
Figure 3.7: Average surface roughness as a function of peak pulse power in fibre laser cutting ................................................................................................. 63
Figure 3.8: Surface roughness, $R_a$ as a function of laser pulse frequency during fibre laser cutting of stainless steel .................................................................63
Figure 3.9: Schematic diagram of the pulse overlap ....................................................64
Figure 3.10: SEM images to demonstrate the comparison of two PO values after one scan, (a) PO=99% and (b) PO=14% .........................................................64
Figure 3.11: Influence of the cutting speed on the kerf width ....................................65
Figure 3.12: Nd:YAG laser cut stents (a) cutting zone, (b) outer surface and (c) inner surface, covered with oxide layer and slag .............................................69
Figure 3.13: Higher average power cause heat effect to the opposite site of the tube. .....................................................................................................................69
Figure 3.14: Prominent recast is found around the channel cut with femtosecond laser fluence as high as 40 J/cm$^2$. .................................................................72
Figure 3.15: Post processing stages after laser cutting process .................................74
Figure 3.16: Stent strut (a) as cut surface (b) acid pickled (c) annealed (d) electropolished .............................................................. .................................75
Figure 3.17: SEM micrograph showing electrochemically polished surface of (a) entire stent (b) cutting zone (c) outer surface and (d) inner surface . ...... 76
Figure 3.18: Diagram of heat-affected zones in biodegradable polymers of CO$_2$ and fs laser machined cuts ............................................................................. 78
Figure 3.19: A medical stent micro machined from a biodegradable polymer using a femtosecond laser (Courtesy of LZH/ Cortronik) .........................................79
Figure 3.20: Features required for an ideal stent cutting system .................................79
Figure 4.1: Swisstec Micro T15 machining system ...................................................82
Figure 4.2: The optical setup for the Libra-S system (the red line denotes the laser beam path) ..................................................................................................85
Figure 4.3: PolyVar-MET microscope ....................................................................86
Figure 4.4: Hitachi (S-3400N) scanning electron microscope .................................87
Figure 4.5: VEECO Wyko NT-1100 Interferometer ................................................ 88
Figure 4.6: MicroXam Interferometer ........................................................................ 89
Figure 4.7: MTS Nano Indenter XP .......................................................................... 90
Figure 5.1: Process window for partial and complete cutting at different ranges of average power and cutting speed ............................................................ 95
Figure 5.2: Scanning electron microscope (SEM) image of the laser cut geometry.......... 96
Figure 5.3: Kerf width as a function of laser cutting parameters in dry and wet cutting (a) peak pulse power (b) frequency (c) pulse width (d) cutting speed ................................................................. 98
Figure 5.4: Surface roughness as a function of laser cutting parameters in dry and wet cutting (a) peak pulse power (b) frequency (c) pulse width (d) cutting speed ................................................................. 99
Figure 5.5: Dross deposition at different cutting speed in different cutting condition. ................................................................................................................ 100
Figure 5.6: SEM images of the backwall (a) dry cutting (b) wet cutting ................. 101
Figure 5.7: White light interferometer (Wyko NT1100) images of surface profiles of the back wall (a) dry cutting (b) wet cutting ......................................................... 101
Figure 5.8: Comparison of thermal effect along the cut kerf at 40 W average power (a) dry cutting (b) wet cutting ............................................................... 102
Figure 5.9: Thermal discoloration in stent cutting before removing the excess material (a) dry cutting (b) wet cutting ............................................................... 102
Figure 5.10: Transverse section of the cut kerf showing the changes of the microstructure associated with HAZ when cutting at high average power (40W) (a) dry cutting (b) wet cutting ............................................................... 103
Figure 5.11: Mechanism of molten material behaviour after ejected from cut kerf in tube cutting in Y-Z view (a) dry cutting (b) wet cutting .............................. 105
Figure 5.12: Water transmission spectrums recorded by spectrometer for 1 cm water path length ........................................................................................................ 106
Figure 5.13: Water absorption spectrums ...................................................................... 107
Figure 6.1: Schematic diagram of laser tube cutting ................................................ 113
Figure 6.2: SEM images of laser cut samples: (a) nitinol and (b) platinum-iridium alloy. ................................................................. 114

Figure 6.3: Nitinol cut edge without any post processing showing the dross free cut and sharp edge in different magnifications: (a) 60x and (b) 200x. ........ 115

Figure 6.4: Platinum-iridium alloy cut edge without any post processing showing the dross free cut and sharp edge in different magnifications (a) 60x (b) 400x. ................................................................. 115

Figure 6.5: Nitinol cut surface tomography: (a) Cut edge, (b) Enlarged part of the analyzed cut edge and (c) Interferometer image giving the surface profiles data. ................................................................. 116

Figure 6.6: Platinum-iridium alloy cut surface tomography: (a) Cut edge, (b) Enlarged part of the analyzed cut edge and (c) Interferometer image giving the surface profiles data. ................................................................. 116

Figure 6.7: 3D image of the nitinol cut surface reconstructed by MeX software package. L1 and L2 represent the path length for striations measurement, (b) Striations profile across the cut edge upper part, L1, (c) Striations profile across the cut edge lower part, L2. ................................................................. 117

Figure 6.8: 3D image of the platinum-iridium alloy cut surface reconstructed by MeX software package. L1 and L2 represent the path length for striations measurement, (b) Striations profile across the cut edge upper part, L1, (c) Striations profile across the cut edge lower part, L2. ................................................................. 118

Figure 6.9: Hardness test at various distances from cut edge for the nitinol sample: (a) measured surface and (b) hardness value. ................................................................. 120

Figure 6.10: Hardness test at various distances from the cut edge for the platinum-iridium alloy sample, (a) measured surface and (b) hardness value. ..... 120

Figure 6.11: SEM micrograph showing inner wall of nitinol is clean from the debris/spatter adherence. ................................................................. 121

Figure 6.12: SEM images showing a great potential of pico second laser in cutting other stent materials, (a) and (b) cobalt chromium, (c) silver, and (d) titanium. ................................................................. 123
Figure 6.13: Picosecond laser benefits in cutting low melting point stent material (polymer)........................................................................................................124

Figure 7.1: The experimental setup used in femtosecond laser machining (the red line denotes the laser beam path). .................................................................132

Figure 7.2: Focal position vs kerf width (13 J/cm$^2$ and 0.1 mm/s). .................133

Figure 7.3: Schematic of the setup used for laser ablated micromachining of nitinol:
(a) dry and (b) underwater..............................................................................135

Figure 7.4: Kerf width as a function of fluence at different cutting speeds for dry and underwater machining: (a) 0.1 mm/s (b) 0.5 mm/s (c) 1.0 mm/s (d) 1.5 mm/s........................................................................................................138

Figure 7.5: Surface roughness, R$_a$ as a function of fluence at different cutting speeds for dry and underwater machining: (a) 0.1 mm/s (b) 0.5 mm/s (c) 1.0 mm/s and (d) 1.5 mm/s. .........................................................................................139

Figure 7.6: Surface tomography at different fluences and the same cutting speed (1.5 mm/s) in dry machining: (a) 4 J/cm$^2$ (b) 6 J/cm$^2$ (c) 8 J/cm$^2$ and (d) 13 J/cm$^2$........................................................................................................140

Figure 7.7: Surface tomography at different fluences at the optimal speed, 0.1 mm/s in underwater machining: (a) 4 J/cm$^2$ (b) 6 J/cm$^2$ (c) 8 J/cm$^2$ and (d) 13 J/cm$^2$........................................................................................................141

Figure 7.8: Nitinol cut surface tomography for dry machining at 4 J/cm$^2$ and 1.5 mm/s: (a) cut edge (b) interferometer image giving the surface profiles data and (c) striations profile across the cut edge. ......................................142

Figure 7.9: Nitinol cut surface tomography for underwater machining at 13 J/cm$^2$ and 0.1 mm/s: (a) cut edge (b) interferometer image giving the surface profiles data and (c) striations profile across the cut edge. .........................143

Figure 7.10: HAZ at different fluences (cutting speed 0.1 mm/s). .......................144

Figure 7.11: (a) Recast formation on the cut edge with dry machining, (b) Recast-free on the cut edge with underwater machining (both samples were processed at 13 J/cm$^2$, 0.1 mm/s)..........................................................................................145

Figure 7.12: Dry cutting showing the debris and the recast formation on the kerf edge. ......................................................................................................................................145
Figure 7.13: Underwater cutting with debris free and recast free result. ...................... 146
Figure 7.14: (a) Debris after dry cutting of nitinol, (b) Debris removed after ultrasonic cleaning, (c) Close up view of the recast after ultrasonic cleaning. ................................................................................................................................. 147
Figure 7.15: (a) Debris-free and recast- free underwater laser cutting does not required ultrasonic cleaning (b) Close up view of the cut. ..................... 147
Figure 7.16: Schematic diagram of interaction between fs laser beam and nitinol in dry machining......................................................................................................................... 148
Figure 7.17: Schematic diagram of interaction between fs laser beam, water and nitinol in underwater machining. ........................................................... 150
Figure 7.18: Example of corrugated surface generated by underwater cutting at high speed, 1.5 mm/s (a)13 J/cm$^2$ and (b) 8 J/cm$^2$. ........................................ 151
Figure 7.19: (a) Debris and recast formation during the cutting process assisting by compressed air (b) close up view. .......................................................... 151
Figure 7.20: (a) Tube backwall was covered by debris in dry machining (b) Clean backwall was obtained in underwater machining. ......................... 153
Figure 8.1: The configuration of the laser and the workpiece. Adapted from Ng. et al. [194]. ...................................................................................................................... 158
Figure 8.2: The physical process in laser micro-machining and drilling process with a static beam, (a) initial arrangement (b) partial penetration depth. ...... 159
Figure 8.3: Mechanism of ejected molten metal behaviour in tube machining, (a) dry machining and (b) wet machining.......................................................... 159
Figure 8.4: Neighbours of particles i within a support kernel. .................................. 165
Figure 8.5: Zoom of a solid surface of a surface particle. The incomplete kernel support of a surface particle gives a non-zero value for divergence $\nabla \cdot \mathbf{r}$. ................................................................................................................... 168
Figure 8.6: Domain for SPH simulation (a) dry cutting (b) wet cutting. ............... 169
Figure 8.7: Cross section of the workpiece showing the phase changes in dry cutting.
Full depth penetration with single pulse (Peak power, $P_p = 100$ W, pulse duration, $\tau = 0.15$ ms). ................................................................. 171
Figure 8.8: Penetration depth evolution as a function of time obtained by the SPH model and experimentally at different pulse durations ($\tau = 0.05$ ms, $\tau = 0.1$ ms and $\tau = 0.15$ ms) with 100 W peak power at a single pulse. ..... 172

Figure 8.9: Penetration depth obtained experimentally at different pulse duration with 100 W peak power at single pulse (a) $\tau = 0.05$ ms, (b) $\tau = 0.1$ ms and (c) $\tau = 0.15$ ms. ............................................................................................ 173

Figure 8.10: Temperature plot indicate the phase change in the material obtained by SPH model obtained from the SPH model. ($P_p = 100$ W, $\tau = 0.15$ ms). 173

Figure 8.11: (a) Average particle velocity in a function material penetration depth obtained by the model for the full penetration depth and (b) Velocity magnitude of the ejected particles. ($P_p = 100$ W, $\tau = 0.15$ ms). ............ 175

Figure 8.12: Example of recast and spatter formation on the cut surface obtained experimentally............................................................. 176

Figure 8.13: Recast and spatter predicted by the SPH model, (a) formation of recast and spatter during the process, (b) recast and spatter after the cutting process..................................................................................................... 177

Figure 8.14: Snapshots of melt ejection time history during wet cutting (water velocity, $V_w = 0.241$ ms$^{-1}$), (a) 0.15 ms (b) 0.175 ms (c) 0.2 ms and (d) 0.25 ms. Note: the particles were coloured by density. ...................... 178

Figure 8.15: Ejected particles behaviour after ejected from the kerf in dry cutting and wet cutting, (a) at $t = 0.15$ ms (soon after ejected from the kerf) and (b) at $t = 0.25$ ms. Note: the particles were coloured by temperature. Cross section of the model is shown in z-y view for a clear capture of the molten materials. .................................................................................................. 179

Figure 8.16: Spatial distribution of the ejected particles deposited to the back wall in SPH simulation (top images) and SEM images zoom of backwall obtained experimentally (bottom images), (a) dry cutting and (b) wet cutting............................................................................................................................................. 180
List of Tables

Table 2.1: Comparison of properties of different inert gas ........................................ 38
Table 3.1: A summary of major stent designs and their manufacturers ..................... 53
Table 3.2: Definition of the “ideal stent” ..................................................................... 54
Table 3.3: A Summary of materials used in balloon-expandable and self-expandable stents, different stent forms, stent fabrication, stent geometry and additions ................................................................. 57
Table 3.4: The advantages and disadvantages of different fabrication methods ...... 58
Table 3.5: Typical cutting parameters by using Nd:YAG lasers summarised from following ................................................................. 68
Table 3.6: Typical laser parameters used in stents cutting and micro-machining for fibre laser summarised from following .................................................. 70
Table 5.1: Chemical composition of stainless steel 316L ........................................ 94
Table 5.2: Cutting parameter ranges ......................................................................... 96
Table 6.1: Process parameters for machining nitinol................................................. 113
Table 6.2: Process parameters for machining platinum-iridium alloy ..................... 113
Table 6.3: Comparison of cutting quality between short and long pulses laser ...... 125
Table 7.1: Material properties of nitinol used in this work ........................................ 134
Table 7.2: Number of passes required to achieve full-depth penetration cut in dry and underwater ......................................................................................... 136
Table 8.1. Thermophysical properties of stainless steel 316 L ............................... 160
Table 8.2: N₂ assist gas and nozzle parameters .......................................................... 160
Table 8.3. Properties of water ................................................................................. 160
Declaration

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Noorhafiza Muhammad, September 2012.
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List of Publications and Award

**Journal papers**


**Conference papers**


**Award**

Nomenclature

\begin{itemize}
\item $A$ absorbance
\item $A_{eff}$ effective area of gas entering the kerf (m$^2$)
\item $A_r$ cylindrical area of radial loss of gas pressure (m$^2$)
\item $c_p$ specific heat (J kg$^{-1}$K$^{-1}$)
\item $d_n$ nozzle exit diameter (m)
\item $f$ frequency (Hz)
\item $g$ gravitational acceleration (m s$^{-2}$)
\item $h_c$ heat convection factor (W m$^{-2}$K$^{-1}$)
\item $h$ smoothing length
\item $i, j, k$ unit vectors
\item $I_i$ Initial laser intensity (W m$^2$)
\item $I_f$ Laser intensity reaching the backwall (W m$^2$)
\item $k_s$ thermal conductivity of solid material (W m$^{-1}$K$^{-1}$)
\item $k_w$ thermal conductivity of water (W m$^{-1}$K$^{-1}$)
\item $L_v$ latent heat of vaporisation (J kg$^{-1}$)
\item $m$ mass of a particle (kg)
\item $n$ number of dimensions
\item $P_{ave}$ laser average power (W)
\item $P_{laser}$ laser power (W)
\item $P_p$ laser peak power (W)
\item $P_0$ atmospheric pressure (Pa)
\end{itemize}
Nomenclature

\( P_c \)  
gas pressure at the nozzle exit (Pa)

\( P_{\text{eff}} \)  
ext择ive gas pressure (Pa)

\( P_i \)  
gas pressure inside the nozzle at position height \( i \) (Pa)

\( P_{\text{vap}} \)  
vapour pressure (Pa)

\( \mathbf{r} \)  
coordinate vector

\( r_d \)  
radial distance from the beam centre (m)

\( r_b \)  
laser beam radius (m)

\( R \)  
gas constant (J kg\(^{-1}\) K\(^{-1}\))

\( t \)  
time (s)

\( T \)  
transmittance

\( T_0 \)  
initial temperature (K)

\( T_b \)  
boiling temperature (K)

\( T_m \)  
melting temperature (K)

\( T_s \)  
melt surface temperature (K)

\( Q \)  
laser source term (W m\(^{-2}\))

\( Q_v \)  
heat loss due to convection and radiation (W m\(^{-2}\))

\( u, v, w \)  
components of velocity vector

\( x \)  
water path length (m)

\( \mathbf{v} \)  
velocity vector (m s\(^{-1}\))

\( V_m \)  
melt ejection velocity (m s\(^{-1}\))

\( W \)  
kernel function

\( z_n \)  
nozzle-workpiece distance (m)

\( \Delta_c \)  
initial particle distance (m)

\( \Pi \)  
artificial viscosity

\( \sigma_b \)  
Stefan-Boltzmann coefficient (W m\(^{-2}\)K\(^{-4}\))
<table>
<thead>
<tr>
<th>Symbol</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>( \varepsilon )</td>
<td>emissivity</td>
</tr>
<tr>
<td>( \rho )</td>
<td>density of a particle (kg m(^{-3}))</td>
</tr>
<tr>
<td>( \rho_m )</td>
<td>melt density (kg m(^{-3}))</td>
</tr>
<tr>
<td>( \rho_s )</td>
<td>density of solid (kg m(^{-3}))</td>
</tr>
<tr>
<td>( \rho_w )</td>
<td>density of water (kg m(^{-3}))</td>
</tr>
<tr>
<td>( \tau )</td>
<td>pulse duration (s)</td>
</tr>
<tr>
<td>( \gamma_g )</td>
<td>gas specific heat ratio</td>
</tr>
</tbody>
</table>
Chapter 1

Introduction

1.1 Research motivation and rationale

A laser is a device capable of amplifying light based on the stimulated emission of high energy photons [1-3]. Laser technology was introduced in industry since the mid-1970s. The explosive growth of laser technology led to the development of lasers from the far infrared down to the x-ray wavelengths. Also pulse duration range has been widened from the millisecond range to the extremely short femtosecond that can be achieved by chirped pulse amplification. Nowadays, lasers are widely used in diverse fields such as material processing, data storage, image recording, printing, defence industry and biomedicine [3, 4].

One of the growing applications of lasers in biomedical engineering is the laser cutting of coronary stents for coronary artery disease treatments. A stent is a wire mesh tube which is deployed in a diseased coronary artery to provide smooth blood circulation [5]. The earliest endovascular stent was pioneered by Dotter and Judkins in 1964 [6]. After twenty two years, in 1986, Jacques Puel implanted a coronary stent
into a patient for the very first time [7]. The methods for producing stents include electrical discharge machining, braiding, knitting, welding, photochemical etching and laser cutting. The first laser cut stent (Palmaz-Schatz stent) for clinical practice was approved by the Food and Drug Administration (FDA) in the United States in 1994 [8]. Over the last decade, most commercial coronary stents are manufactured by laser cutting of tubular materials [9].

The suitability and economical benefits of using laser cutting technology for stent fabrication has been extensively proven [10]. Starting with the application of lamp-pumped Nd:YAG lasers (wavelength of 1064 nm and pulse duration in the millisecond range), laser technology has developed quickly to suit the demands in cutting small medical devices by ultra-short laser pulses of only a few femtosecond. The advantage of using lasers against conventional methods is the ability to manufacture stents with the highest speed, precision and quality in order to meet the stringent requirements for implantation in the human body.

There are, however, still many challenges in laser cutting of stents. Inconsistency in the geometry of stent struts can result in an inaccurate stent expansion index when implanted in the human body. Thermal distortion inherently creates a heat affected zone (HAZ) in the vicinity of the cut region that could result in cracking when the stent is expanded due to material embrittlement. A high surface quality cut with no trace of dross, debris and recast layer has to be achieved for a good quality stent. The other issue in cutting tubular material is backwall damage. Nevertheless, these quality issues of laser cutting of stents can be addressed by post processing. Depending on the quality obtained after the cutting process, multiple post processing methods such as soft etching, acid pickling, electropolishing can be undertaken to produce stents that meet the industrial standard [8, 11].

A recent report from Nursing and Health Studies (NHS) [12] states that one in every four men and one in every six women die and approximately 300,000 people suffering heart attack each year in the United Kingdom. American Heart Association reported that coronary artery disease killed about 650,000 American annually [13].
More than 3 million stents are implanted worldwide every year [14]. In the United States, 600,000 stents are implanted each year and in the United Kingdom, the latest figure indicates that around 50,000 coronary stenting are implanted each year [15]. Each bare metal stent cost about £500 while drug eluting stent cost about £2000 each [16].

With the demands of coronary stents rising every year, the stent production demands a higher quality, lower cost and faster laser cutting process that requires a minimum number of post processing steps. Quality improvement in laser micromachining of tubular materials for coronary stents is the main factor that has driven research development in the present PhD project. This research project has been carried out utilising three different lasers: fibre laser, picosecond laser and femtosecond laser with different processing materials (i.e. stainless steel 316L, nitinol, titanium, cobalt chromium, silver and polymer) and methods (gas and water assisted cutting).

## 1.2 Aim and objectives

### 1.2.1 Aim

The aim of this project is to further advance laser micromachining of biocompatible materials for coronary stent applications and extend the understanding of the processes. It is intended to develop a process capable of achieving high quality cutting by studying different types of lasers, materials and methods for cut quality improvement.

### 1.2.2 Objectives

The objectives of this research are summarised as follows:
Chapter 1: Introduction

- To investigate the basic characteristics of water assisted fibre laser micromachining of stainless steel 316L tubes and to understand the effect of introducing water flow in the process to minimise back wall damage and thermal effects.

- To demonstrate the capability and to understand the process characteristics of picosecond laser machining of nitinol, platinum iridium alloy and other stent materials.

- To investigate and compare the quality characteristics between dry and underwater cutting of nitinol tubes using a femtosecond laser.

- To further understand the process characteristics of pulsed laser dry and wet micromachining processes by multi-phase smoothed particle hydrodynamics (SPH) modelling method.

1.3 Thesis outline

Chapter 2 reviews the laser fundamentals and its capabilities. The advantages of the laser cutting process are described. The basic laser cutting mechanisms are also reviewed including the key process parameters and characteristics of laser cutting.

Chapter 3 provides a literature review on stent manufacture with a particular focus on the characteristics of laser micromachining of stents. The general background of coronary stents is presented along with the historical development of stent cutting and current status to identify the knowledge gap and the need for further investigation.
Chapter 4 describes the experimental equipment employed in this research work including the stent cutting platform, laser systems, analytical equipment and material characterisation methods.

Chapter 5 reports on the experimental work carried out to compare between dry and wet fibre laser cutting of stainless steel 316L tubes. The effect of introducing water flow in the processing of tubes to minimise backwall damage and thermal effects was investigated. In addition, the influence of laser parameters upon cutting quality is discussed.

Chapter 6 details the work done on picosecond laser micromachining with special emphasis on cutting nitinol and platinum-iridium alloys tubes. Using the third harmonic laser radiation of a picosecond laser at 6 ps pulse duration in the UV range, the capability and the characteristics of the picosecond laser in profiling nitinol and platinum-iridium alloys are presented. It also discusses the potential of picosecond lasers in machining other stent materials such as cobalt chromium, silver, titanium and polymer.

Chapter 7 presents research work in the femtosecond laser micromachining of nitinol inside a confined, thin water layer compared to processing in air. The characteristics of laser interaction with the material are discussed. The advantages of water assisted cutting are described and the physical processes involved during cutting through a thin water film (i.e. bubble formation and shock wave) are also discussed.

Chapter 8 presents the SPH modelling of laser cutting of thin materials. A three dimensional SPH modelling method is utilised to develop the numerical model. The open source code SPHysics is used to model the interaction between laser beam and workpiece. This chapter demonstrated that the meshless characteristics of SPH are able to model the melt splashing and droplets leaving the kerf, which is difficult to achieve by conventional modelling techniques. Water is also included in the model to help explain the wet cutting mechanism in laser cutting.
Chapter 9 presents the conclusions of this research work and recommendations for future research.
Chapter 2

Review of Laser Cutting Fundamentals

2.1 Introduction

This chapter introduces laser cutting fundamentals in order to understand the basic process characteristics and material removal mechanisms. The important parameters and quality characteristics typically considered in laser cutting are discussed. Types of lasers used in laser cutting were also introduced.

2.2 Laser cutting introduction

The history of laser began when Albert Einstein introduced the theory of stimulated light emission back in 1917. Following Einstein’s theory, Theodore Maiman demonstrated the first ruby laser in 1960 [17]. Since then, research in lasers has progressed with the development of different types of lasers for various applications such as laser cutting, welding, drilling and surface engineering.
Laser cutting is a process in which a high intensity laser beam is used to rapidly heat the target, subsequently melting and/or vaporising the material through full depth of the target material. The generated molten material and vapour is blown out from the cut kerf by an assist gas jet [18]. The use of oxygen as an assist gas jet potentially increases the efficiency of the process due to the exothermic reaction (provide additional thermal input) with certain materials at high temperatures to add additional exothermic energy. For example cutting mild steels in the presence of pure oxygen as an assist gas can increase the cutting speed of the process by 50% compared to that using an inert gas [17, 19]. Figure 2.1 illustrates the basic configuration of a laser cutting process. The laser beam is focussed/ narrowed by a focussing lens down to a smallest spot. Small spot size is preferred during the cutting process for a precise material removal and high accuracy. Coaxial gas is delivered through a nozzle in the same direction with the laser beam. The introduction of assist gas is to enhance the ejection of melt/ vaporised materials out of the kerf. This also helps to prevent the optics from the melts splashing which will damage the optics. Nozzle-workpiece distance is also important during the cutting process. A slight change of the nozzle distance significantly affects the cut quality. The typical distance is usually maintained around 0.5 – 1.5 mm, where good cut quality can be obtained, depending on the process [1, 20].

![Figure 2.1: Schematic diagram of the laser cutting process](image-url)
There are different types of continuous wave and pulsed lasers used in cutting process include CO\textsubscript{2} laser, Nd:YAG laser, fibre laser and ultra-short pulse laser. Wide range of materials can be cut by lasers such as pure metal, metal alloys, wood, paper, ceramics, polymers, leather and acrylics. The applications of laser cutting are numerous including metal cutting for automotive and aerospace components, cutting of biocompatible materials for medical device applications, cloth cutting, die board cutting for electronic applications, cutting of quartz tubes for car halogen lamps, Kevlar cutting and radioactive materials cutting [1, 2, 22].

2.2.1 The advantages of laser cutting

Laser cutting offers a number of advantageous features over some machining techniques as enlisted below [1, 2, 18, 19, 22-30]:

- It is a non-contact process and it does not require the clamping of the workpiece avoiding the mechanical damage from the clamping (due to none physical contact) and avoidance of tool wear and associated costs.

- Ease of automation with computer numerical control (CNC) and robotic processing, which provides accurate control over the process (i.e. dimensions and speeds) suitable for cutting complex shapes.

- Small spot size (down to a few micron) that offers high geometry accuracy (down to a few µm).

- High cutting speed that improves productivity.

- Almost all types of materials can be cut by a laser beam.

- Minimal heat affected zone (HAZ) – compared with other thermal cutting processes such as plasma, flame etc.

- A quiet process (noise level reduction at the workplace).
2.2.2 Laser cutting mechanisms

Different laser cutting approaches can be obtained based on the interaction between the laser beam and the materials also by using different type of assist gases which have a significant effect in the material removal mechanism [18]. The main laser cutting mechanisms are based on fusion cutting, reactive fusion cutting and vaporisation cutting, which are defined according to the material removal mechanisms [18, 22] as shown in Figure 2.2.

![Laser cutting approaches](image)

Figure 2.2: Laser cutting approaches: (a) fusion cutting, (b) reactive fusion cutting and (c) vaporisation cutting [29].

2.2.2.1 Fusion cutting

Laser fusion cutting involves a melting mechanism where the heated materials transform into a molten state and then expelled from the cut kerf by a high pressure assist gas jet (see Figure 2.2a). Inert gases (i.e. nitrogen, argon and helium) are generally used during the process to blow out the melted material, to shield the cutting point from the surrounding air, to avoid sideways burning and to minimise dross attachment at the bottom edges of the kerf. Dross adherence occurs when the supplied gas jet forces do not exceed the surface tensional force. This cutting approach is applicable to various metals such as stainless steel, titanium alloys and other high alloyed steels, and aluminium alloys [1, 18, 22, 28, 29, 31-33].
2.2.2.2 Reactive fusion cutting

Reactive fusion cutting is similar to the fusion cutting mechanism. However, here oxygen (reactive gas) is used as the assisting gas (see Figure 2.2b). The use of oxygen causes exothermic reactions, which provides additional energy to the process. Therefore, it offers a higher cutting speed compared to the laser fusion cutting (where the energy comes solely from the laser beam). However, this oxidation also promotes the growth of an oxide layer on the cut kerf’s walls due to the diffusion of oxygen atoms in the materials. It also introduces sideways burning to the process. These can be regarded as disadvantages of reactive fusion cutting compared to the fusion cutting. Typical materials that are cut by reactive fusion cutting processes are normally mild steel and low alloyed steel [22, 29, 34, 35].

2.2.2.3 Vaporisation (sublimation) cutting

Vaporisation cutting removes the material mainly in the form of vapour, where the laser provides relatively high energy until the material reaches its vaporisation temperature in a very short interaction time, which is commonly achieved with a pulsed laser beam (see Figure 2.2c). Direct phase transition of the materials to vapour state produces extremely high cut quality with smooth surfaces and clean cut edges. Generally, inert gas is used in order to avoid the oxidation of the cut edges; however, in some cases the assist gas is not required. This method is favourable for cutting materials with low thermal conductivity and low heat of vaporisation such as organic materials, paper, cloth and polymers [18, 29, 34, 35].

2.2.3 Laser cutting process parameters

Laser cutting processes are governed by various processing parameters including laser power and intensity, focal position, pulse duration, cutting speed, repetition
rate, and assist gas pressure and type. The parameter selection must be clearly understood for different lasers and different materials since it will significantly affect the final cutting results in terms of geometric and metallurgical quality [18]. This section discusses the effects of some of the most important laser processing parameters on the geometric and material quality of the cut. A details explanation of the main cutting parameters is attempted in the following paragraphs.

2.2.3.1 Laser power and intensity

Laser power is the energy (in Joules) delivered per second, commonly expressed in Watt. The power intensity of the laser is the power delivered per unit area and can be obtained by dividing the laser power over the concentrated area of the active beam (spot size area). Proper laser beam focusing is very important to obtain high intensity which is desirable in cutting applications. Furthermore, a high laser power allows one to perform the cutting process at higher speeds. A proper selection of laser power has to be established, as too higher laser power (excessive) results in a wider kerf and a thick recast layer. On the other hand, insufficient laser power will produce dross due to incomplete melting [33, 36].

2.2.3.2 Focal position

Variation of the focal position with the same laser beam conditions produces significantly different final cut quality. The focal position is determined by the appropriate alignment of the focus spot size (power intensity) at the target surface or slightly below the workpiece surface for optimum cutting. Different cutting conditions require different locations of the focal spot either on the top of the workpiece, within the workpiece or below the workpiece. Steen [1] stated that an optimal cutting operation can be achieved by positioning the focus just below the top surface. For laser fusion cutting, the focal position is within the workpiece but close to the bottom to minimise dross formation. It has been also found that placing the
focal point at the middle of the workpiece maximises the speed of the cut process [22]. In reactive fusion cutting, it is suggested that placing the focal point within the upper half of the material thickness provides optimum cutting quality. In laser cutting of thicker materials >10 mm, the focal position is usually placed above the workpiece surface [22]. Cutting thicker material involves large volume of material removal, when wider channel is needed. The adjustment of the focal position is importance in increasing the depth of cut. Focal position will significantly affect the cut quality. This is due to the fact that by changing the focal position during the process, the spot diameter, energy distribution and laser beam coupling with the workpiece are also changed [18, 37].

2.2.3.3 Pulse duration

Continuous wave laser cutting is used commonly for most cutting processes. However, in certain applications, particularly for micro-cutting of thin section materials, a pulsed laser beam is used. Pulse duration is an important parameter in laser cutting because it controls the beam/material interaction time and the energy delivered into the workpiece. In today’s laser technology, the pulse duration is available from millisecond down to a few femtosecond for a laser cutting process. Pulse duration indicates the beam and material interaction time and it is one of the most significant parameters related to the HAZ formation. For example, at femtosecond pulse duration the HAZ formation is almost absent [38].

2.2.3.4 Cutting speed and repetition rate

Cutting speed has a significant influence in laser cutting process because it regulates the interaction time between the material and the laser beam. Cutting speed is commonly represented in units of mm/min, mm/s, or m/min. A low cutting speed generally promotes dross formation during fusion cutting and generally causes sideways burning in reactive fusion cutting. It has been found that increasing the
cutting speed can produce a narrow kerf and reduce the HAZ [39]. This can be explained by the short beam and material interaction time at the high speed. Rao et al. [31] stated that the maximum cutting speed (at full depth penetration cut) is one of the most significant parameter in the cutting process in order to obtain high quality cut as well as high throughput. The maximum cutting speed which yields a full depth penetration in the laser cutting process is power dependant [40]. Repetition rate represents number of pulses delivered within a second (expressed in Hz). This is an important process parameter related to the surface roughness and material removal. A high repetition rate provides high pulse overlapping and thus gives a high material removal rate and a finer cut surface [18].

2.2.3.5 Assist gas

The selection of the appropriate assist gas for a laser cutting process is important because it modifies the yield and quality of the cutting operation. Three commonly used assist gas in laser cutting of metallic materials are reactive gas (oxygen), inert gas (nitrogen, argon and helium) and air. The reaction between workpiece materials with the assist gas element has to be understood to take advantage of the extra input energy from exothermic reactions and to avoid the oxidation side effects. Oxygen is the most common assist gas used in laser cutting mild steels. Here, an exothermic reaction occurs with oxygen where additional heat is supplied in the cutting process which allows cutting at higher speed. However, the disadvantages of cutting with oxygen are related to the development of oxidation which significantly affects the surface quality. Figure 2.3 shows the acceptable quality region in cutting 3 mm mild steel in CO₂ laser cutting with the presence of oxygen as an assist gas [41].
Inert gas is a favourable gas, especially in cutting stainless steel and titanium in order to avoid oxidation of the materials. Cutting with an inert gas provides a clean cut edge and no oxidation as the energy is solely from the laser beam and the assist is only used for ejecting the molten metals. High pressure inert gas (up to 14 bar) is used during the cutting process to enhance drag forces and achieve high cut quality [1]. Rao et al. [31] performed laser cutting of titanium with different assist gases Argon (Ar), Helium (He) and Nitrogen (N$_2$) to compare the effects of these gases upon cut quality. Among these three gases, it was found that the use of He as an assist gas showed more significant effect in dross reduction as compared to Ar and N$_2$. This can be explained due to the high shear stress produced by He as can be seen in Table 2.1.
2.2.4 Characteristics of laser cut materials

The quality of a laser cutting process can be assessed by various material characteristics including kerf width, surface roughness, heat affected zone (HAZ), dross, recast and striations formation [40].

2.2.4.1 Kerf width

Kerf width is a terminology used to define the distance between the edges of the laser cut slot [42]. A narrow kerf width is favourable in laser cutting as this indicates a precise material removal process (minimum amount of waste material) [22]. In most cases, the top kerf width is found slightly wider as compared to the bottom kerf, forming a taper cut nature. The occurrence of this phenomenon can be explained due to the several factors, i.e. beam intensity losses, beam defocusing and gas pressure losses across the material thickness [40]. The laser power, cutting speed and assist gas jet have been found to have major influence on the size of the kerf width [43].
2.2.4.2 Dross

Dross is a molten material and witnessed as a solidified drops clinging to the bottom edges of the cut materials. Surface tension and viscosity of the molten material influence significantly the formation of the dross. High surface tension and viscosity resist the molten metal from flowing smoothly out of the cutting zone and it develops the dross formation as shown in Figure 2.4 [22]. Assist gas jet is utilised to expel the molten material during laser cutting. Although these molten materials are not completely removed, the assist gas jet is able to minimise the dross adherence. In reactive fusion cutting, dross is identified to occur when a low gas pressure is used during the process. Other reasons include improper focus and high cutting speed. Furthermore, laser cutting with an inert gas jet increases the chances of dross occurrence compared to laser cutting with reactive gas. This is due to the fact that surface tension of pure metals is higher than oxidised metals. The approach to achieve dross-free cutting with an assist inert gas is by increasing its pressure up to 10 bar or more [22, 44].

![Dross formation](image)

Figure 2.4: Dross formation [45].

2.2.4.3 Heat Affected Zone (HAZ)

The HAZ is an acknowledgement of the laser cutting process being a thermal phenomenon where a huge amount of energy is absorbed and conducted into the workpiece. The excessive heat input above the required energy to reach the phase
transformation temperature of the material processed changes its properties and alters the microstructure of the workpiece [46]. HAZ can be found in the vicinity of the laser cut area that experiences excessive heat but it does not reach the melting point. This region is heat distorted compared to the parent material and usually cracks are observed. The HAZ width can be measured from the beginning of the cut edge to the perpendicular direction where the HAZ can be identified by observing the microstructure or hardness variation adjacent to the cut edge [46-48].

### 2.2.4.4 Recast

Recast is formed due to the re-deposition of the molten materials on the cut edges. This layer is usually harder than the parent materials, and also is highly stressed which may lead to crack formation as shown in Figure 2.5 [49]. Lv et al. [49] found that laser parameters (pulse energy, assist gas and cutting speed) highly influence in the formation of recast layer. The increasing of laser pulse energy resulted in significant increase in the energy delivered to the process, resulting in a thicker recast layer. Recast layer can be reduced by increasing the cutting speed where reduction of energy density on the cut area can be expected. A high pressure inert gas (10 bar and greater) is found to reduce recast layer. Cutting with ultra-short laser pulses (pulse durations of $10^{-12}$ - $10^{-15}$ s) were also suggested to potentially reduce recast thickness [11].

Figure 2.5: (a) SEM photograph and hardness variation of the transverse section of a typical laser cut surface [49].
2.2.4.5 Surface roughness

Surface roughness characterises the texture and quality of the cut surface. It appears in the form of irregularities of the surface profile. Several different profile roughness parameters are used to represent the roughness quality. However, average roughness, $R_a$, is the most commonly used parameter for the quantitative representation of surface finishes. Other profile parameters are the maximum peak height, $R_p$, and maximum valley depth, $R_v$. Figure 2.6 illustrates five single profile element where the average roughness of the profile can be determined [20, 36, 42].

![Diagram](image)

Figure 2.6: Mean height of the profile [42].

2.2.4.6 Striations

Striations are defined as periodic lines with a slight inclination from the beam axis appearing on the cut edge after the laser cutting process (Figure 2.7). Striations occurrence deteriorates the cut edges and cause uneven surface roughness. It is reported that more than one striation patterns can be observed on the cut edges [47, 50]. A considerable amount of research work has been conducted to explain and to gain a better understanding of the striations development phenomena. Researchers have inferred that striations are formed due to sideways burning [51], cyclic
oxidation reaction [52], hydrodynamic instability [53] and temperature fluctuation during the process [54]. Striations-free cuts can be obtained by a proper selection of process parameters. Sobih et al. [50] presented an operating window which produced striations-free cuts in laser cutting of 1 and 2 mm mild steel sheets using oxygen as assisting gas jet. A mathematical model can be found in a work by Li et al. [55] which predicts the critical cutting speed for striations-free cutting. Recent work by Yan et al. [56] demonstrated the optimal operation window to achieve striations-free cutting of 1 mm alumina by using a 400 W nano-second pulsed DPSS Nd:YAG laser. They have applied the theoretical model developed previously by Li et al. [55] to predict the cutting speed where the striations-free cutting can be achieved.

Figure 2.7: Striations appear in the form of periodic lines [52].

2.3 Types of lasers

There are different types of lasers used in material processing including carbon dioxide (CO₂) lasers, neodymium-doped yttrium aluminium garnet (Nd:YAG) lasers, ytterbium-doped fibre lasers, excimer (KrF, ArF, XeCl) lasers and Ti:Sapphire lasers [1]. In laser cutting and micromachining, CO₂, Nd:YAG, fibre and ultra-short pulse
lasers (Ti:Sapphire) are the most common lasers applied. These lasers are summarised in the following sections.

2.3.1 CO$_2$ laser

CO$_2$ lasers are one of the molecular gas lasers, which are one of the earliest lasers introduced in industrial application. This laser emits at a wavelength of 10.6 $\mu$m (infrared spectrum), CO$_2$ lasers contain a mixture of CO$_2$ (laser active molecules), N$_2$ and He. The role of N$_2$ molecules in the mixture is to transfer their energy to CO$_2$ molecules by collision with electrons in the discharge. The presence of He as an internal heat sink (absorb heat) facilitates CO$_2$ molecules to get back to the ground level where CO$_2$ and N$_2$ repeatedly interact for an ongoing lasing process. A CO$_2$ laser can be operated in two modes: continuous wave (CW) and pulsed mode. The power for this laser can be within a range of a few mW up to tens of kW. Different designs of CO$_2$ laser available include: transverse flow (for powers greater than 4 kW, usually with a poor beam quality), fast-axial flow laser (for powers between 500 W to 8 kW, and is the most common laser for cutting applications), slow axial flow laser (for powers less than 1 kW), diffusion cooled slab laser (high energy efficiency, up to 30 kW), and sealed-off laser (for powers less than 1 kW). Metallic materials have a very low absorption towards CO$_2$ laser beam but the beam is highly absorbed by non-metallic materials (ceramics or organic materials) [17, 27, 36, 57, 58].

2.3.2 Nd:YAG laser

A Nd:YAG laser is a solid state laser with a wavelength of 1.06 $\mu$m (near infrared). The lasing material is an yttrium-aluminium garnet (YAG, chemical formula is $Y_3Al_5O_{12}$) doped with neodymium (Nd$^{3+}$ ion). This laser can be operated in two modes: continuous wave (cw) and pulsed mode. For pulsed mode, the pulse duration can be within milliseconds up to microsecond range with high peak powers (up to tens of kW) and repetition rate of several kHz. Recent development of Q-switched
and mode locked Nd:YAG lasers allow the pulse length to be extended to nanosecond and picosecond. Flash lamp pumped Nd:YAG lasers are the primary laser source that was used in micromachining (stent manufacturing) at the beginning of applications. Apart from cutting and micromachining, the Nd:YAG lasers are also used for welding and hole piercing [27, 36, 59]. The laser beam can be transmitted and delivered via optical fibre which increases the process flexibility. On the other hand, due to shorter wavelength, Nd:YAG laser is used in machining high reflective material (metallic materials) which are difficult to process by CO₂ lasers. However, one of the drawbacks of this laser is that it requires higher running and capital cost as compared to a CO₂ laser [17, 57].

2.3.3 Fibre laser

A fibre laser in the realm of laser material processing is regarded as the new development for solid state laser technology and it is extensively used in many industrial applications. It was first invented by Elias Snitzer in 1963 with a power in milliwatt range [60]. The modern fibre lasers can operate up to 100 kW.

The configuration of a fibre laser is as follows: the lasing medium is a silica fibre doped with a rare earth element (Ytterbium, Erbium or Thulium) which is optically pumped by a diode laser (laser diodes, diodes stalks and diodes emitters) at the ends and through the cladded shell surface (consisting of inner and low-refraction outer cladding) as illustrated in Figure 2.8.
Figure 2.8: A schematic drawing of the double-clad fibre [61].

The wavelength generated by a fibre laser is within the range from 1060 to 1085 nm. A fibre laser consists of a long fibre laser cavity of approximately 12 metres in which the pump energy is deployed along the full length [61-65]. The fibre lasers can be excited by two methods, end-pumped and multi-mode pumped. With the end-pumped excitation method, the laser light pumped by diode bar is focussed to the sample by a set of lens which is guided by a fibre. The arrangement requires the diode bar, focussing lens, and fibre to be constantly aligned correctly [66]. A typical configuration of end-pumped fibre laser is shown in Figure 2.9.

Figure 2.9: Configuration of end pumped fibre laser [63, 65].
Figure 2.10 shows the configuration of multi-mode pumped fibre laser. Laser light is pumped by a stack of diodes emitters [65]. In this case, the resonator is formed by Bragg gratings that require no mirrors or lenses at the end-pumped fibre laser. Thus, there is no requirement for cooling, cleaning or alignment needed. It is possible to enhance the quality of the laser beam with the use of finer fibres [66]. This laser is less complex in its construction compared to the aforementioned excitation method. However, it requires more advanced technologies for production [64-66].

![Figure 2.10: Configuration of multi-mode pumped fibre laser [63].](image)

Fibre lasers have various advantages over other solid state lasers. Among them is its superior beam quality which gives an excellent focal performance. The beam quality, $M^2$ can be expressed as [63]:

$$M^2 = \frac{BPP \pi}{\lambda}$$  \hspace{1cm} (2.1)

$$BPP = \alpha \omega_0$$  \hspace{1cm} (2.2)

Where BPP is the beam parameter product, $\lambda$ is the laser wavelength, $\alpha$ is the beam divergence, and $\omega_0$ is the beam spot radius. The beam quality with a value of 1 is regarded as a superior Gaussian beam quality whilst higher than 1 is commonly seen
for commercial fibre lasers. These two parameters; beam quality factor and beam parameter product are useful for comparing different lasers with same wavelength [61]. Fibre lasers are reliable laser sources with high operating efficiency, inherently simple, excellent beam parameter product (as compared to the traditional sources), large depth of focus, smaller focussed spot diameter, excellent focus ability, higher absorption of radiation for metallic materials, low operating cost, minimal space requirements and small heat exchanger. Those listed advantages easily fulfil requirements in laser material processing specifically in laser cutting [29, 42, 43, 47, 48, 50, 61, 67-69].

Fibre lasers have been found favourable in stent cutting after the traditional solid state laser (Nd:YAG). The review of fibre laser applications in stent cutting will be presented in detail in Chapter 3.

### 2.3.4 Ultra short pulse lasers

The introduction of the ultra-short pulse lasers is a breakthrough in the laser industry. Here, the use of a special amplification technique enables the pulse duration to be compressed to less than picosecond time regime. For instance, the chirped pulse amplification (CPA) is a technique where the laser pulse is stretched before the amplification step [70-72]. The pulse length ranges for ultra-short pulsed lasers are within picosecond to femtosecond range ($10^{-12}$ - $10^{-15}$ s).

The CPA technique comprises a stretcher-compressor system which is based on the generation of an ultra-short pulse from the seed laser. This laser pulse is stretched by a factor of $10^3$ to $100^3$ by a dispersive element (stretcher) to lower its peak power. The stretched pulse (low-brightness pulse) is then amplified and recompressed to its original pulse duration. Figure 2.11 shows the CPA technique for ultra-short pulse laser [71, 73].
Despite the high peak power, the CPA technique enables achieving a very short pulse duration which is suitable for micromachining operations (such as the fabrication of miniature devices). This ultra-short pulse laser has a potential to process stents and other sensitive and delicate materials with minimum thermal effects and high accuracy [74, 75]. The review of ultra-short pulse laser (picosecond and femtosecond) applications in stent cutting will be presented in detail in Chapter 3.

### 2.4 Summary

The fundamentals of the laser cutting technology are reviewed. Different laser cutting mechanisms were discussed as well as the typical processing parameters involved. The fundamentals of the CO₂ laser, Nd:YAG laser, fibre laser and the CPA technique for ultra-short pulse laser generation were also discussed. In laser cutting, the final cut qualities were mainly influenced by the following process parameters: laser power (W), focal position (m), pulse duration (s), cutting speed (m/s), repetition rate (Hz), and type and pressure of the assist gas.
Chapter 3

Laser cutting of coronary stents for cardiovascular treatments – process characteristics and challenges

3.1 Introduction

A coronary stent is a mechanical device designed to open the arteries that have been occluded, this is known as coronary artery disease (CAD). A recent report from Nursing and Health Studies (NHS) [12] states that one in every four men and one in every six women dying from the disease with approximation of 300,000 people suffering heart attack each year in the United Kingdom. American Heart Association reported that coronary artery disease killed about 650,000 Americans annually [13]. More than 3 million stents are implanted worldwide every year [76]. In the United States, 600,000 stents are implanted each year and in the United Kingdom, the latest figure indicates that around 50,000 coronary stenting are performed each year [15]. Coronary artery disease is a big killer worldwide. The stringent design requirement
of coronary stent demands high precision technologies to perform fabrication of mesh-wire like structures.

The stents were conventionally fabricated by electric discharge machining, photochemical etching, braiding and knitting methods. The emergence of laser processing has dominated the stent manufacturing over the last decade due to their high speed, high precision, low cost and reliability. The typical size of stent used in clinical practice are 2.5 to 4.0 mm in diameter and 8 to 38 mm in length to fit the various sizes of diseases artery [77]. They are of wire-mesh structures with a typical wall thickness of 80-100 \( \mu \)m and the strut width of 80 \( \mu \)m. According to the studies carried out by Kastrati et al. [78], thinner strut stents are preferable due to lower profiles yet minimise restenosis after implantation. High accuracy is therefore necessary for stent manufacture and the dimensional accuracy is a critical factor [48].

Laser cutting is one of the key fabrication technologies applied to coronary stent manufacture. This chapter provides a review on recent process and state of the art in laser-based stent manufacturing. This includes an introduction to coronary stents, their basic characteristics and material properties, followed by various fabrication processes. The characteristics of long pulse (millisecond pulse) laser cutting and recent ultrafast (picosecond and femtosecond pulsed) laser micro-fabrication for stent manufacture is reviewed. Different approaches/techniques to aid cut quality improvement are presented. Post-processing stages after laser cutting process are described followed by an outlook of future prospects. This review is aimed to provide an understanding of fundamental science behind the laser fabrication of coronary stents.

### 3.2 Coronary stents

Blood vessel obstruction is the result of atherosclerosis caused by the build up of smooth muscle cells, lipid containing vacuoles, calcifications and additional cellular debris [12]. Stent devices are introduced in clinical uses not to remove the
obstruction but to create a larger lumen within the lesion to provide smooth blood circulation [5]. A stent is a wire mesh tube and it could be either balloon expandable or self-expanding. A balloon-expandable stent is deployed in a coronary artery by mounting it on a balloon catheter, positioning in a blood vessel and expanding it by inflating the balloon. The balloon is then deflated and removed, permanently leaving the stent in place [79]. Self-expandable stents are manufactured at the vessel diameter (or slightly larger). They are crimped and constrained to a small diameter until reaching the intended delivery site, where the constraint is then removed and the stent is deployed [80]. Stent geometry include rings, helix, coil, unconnected, open cell, closed cell, peak-peak, peak-valley and hybrid [9]. Figure 3.1 shows a disease artery (plaque build up) and coronary artery stent placement [81].

Figure 3.1: An illustration of blood vessel blockage and coronary stent placement. (a) An artery (with plaque build up) inserted with a crimped stent mounted on a balloon catheter (b) The balloon is inflated and widening the artery and (c) Widened-artery restored normal blood flow (Photo from National Heart Lung and Blood Institute [81].
Intravascular stenting was pioneered by Dr. Charles Dotter in 1983 who introduced stents for implantation in dog’s femoral artery [82]. Paul and Sigwart deployed the first stent in human coronary artery in 1986 [83]. By 1999, stenting comprised 84.2% of percutaneous coronary inventions [83]. In 2002, a review by Stoeckel et al. [9] approximated over 100 different stent designs for vascular and non-vascular indications. Fundamental understanding of stent technology i.e. shape, thickness, material selection is essential in their appropriate implementations in practice [5].

### 3.3 Bare Metal Stents (BMS) and Drug Eluting Stents (DES)

There are two types of stents currently used in clinical practice; bare metal stents and drug eluting stents. Bare metal stents are metal stents without any anti-proliferative drug coatings. Drug-eluting stents consist of a platform (the stent), a coating material (typically polymer), and a pharmacologic agent (a drug) [83]. The coating material holds and releases the drug to the artery wall. The purpose of the drug is to inhibit restenosis in the artery. Schematic of BMS and DES are shown in Figure 3.2.

Figure 3.2: Schematic of Bare Metal Stent and TAXUS stent (Drug-eluting stent by Boston Scientific) [84].
In 2003, Food and Drug Association (FDA) approved Cordis sirolimus-eluting Cypher stent for use in medical application. Boston Scientific’s paclitaxel-eluting Taxus stent was approved for use in 2004. The superiority of drug-eluting stents highly met the anticipation by the clinical community and within a few years, it was reported that several millions of drug-eluting stents were implanted all over the world [85, 86]. The major stent designs from different stent manufacturer approved by Food and Drug Administration (FDA) for clinical use are listed in Table 3.1.

Table 3.1: A summary of major stent designs and their manufacturers [87-91].

<table>
<thead>
<tr>
<th>MANUFACTURER</th>
<th>REGISTERED TRADEMARKS</th>
<th>BARE METAL STENT</th>
<th>DRUG ELUTING STENT</th>
</tr>
</thead>
<tbody>
<tr>
<td>Abbott Vascular</td>
<td>Vision, Xience V&lt;sup&gt;™&lt;/sup&gt;, Xience Prime&lt;sup&gt;™&lt;/sup&gt;, Multi-Link Vision ST stents</td>
<td>MULTILINK II Coronary Stent System</td>
<td>Multi-Link Visi&lt;br&gt;&lt;br&gt;l Coronary Stent System</td>
</tr>
<tr>
<td>Boston Scientific</td>
<td>OMEGA&lt;sup&gt;™&lt;/sup&gt;, PROMUS&lt;sup&gt;™&lt;/sup&gt;, TAxUS&lt;sub&gt;©&lt;/sub&gt;, TAxUS Express, and TAxUS Liberté&lt;sup&gt;™&lt;/sup&gt; stents</td>
<td>OMEGA&lt;sup&gt;™&lt;/sup&gt; Platinum Chromium Coronary Stent System</td>
<td>PROMUS™ Everolimus-Euting Coronary Stent System</td>
</tr>
<tr>
<td>Cordis Corporation/Johnson &amp; Johnson</td>
<td>Cypher&lt;sup&gt;™&lt;/sup&gt;, Cypher Select&lt;sup&gt;™&lt;/sup&gt; stents</td>
<td>Palmez-Schatz stent</td>
<td>Cyphers&lt;sup&gt;™&lt;/sup&gt; Sirolimus Euting Stent</td>
</tr>
<tr>
<td>Medtronic Vascular</td>
<td>Endeavor&lt;sup&gt;™&lt;/sup&gt;, Driver&lt;sup&gt;™&lt;/sup&gt; stents</td>
<td>Driver Bare-Metal Stent</td>
<td>Endeavor&lt;sup&gt;™&lt;/sup&gt; Drug-Euting Coronary Stent</td>
</tr>
</tbody>
</table>

3.4 Stent Materials

Materials for metallic balloon-expandable or self expanding stents must exhibit excellent corrosion resistance and biocompatibility. They should be sufficiently radiopaque and create minimal artifacts during magnetic resonance imaging (MRI) scan which is usually performed to diagnose health conditions [9]. Momma et al. [71] listed the ideal stent properties required as shown in Table 3.2.
Table 3.2: Definition of the “ideal stent” by Momma et al. [71].

<table>
<thead>
<tr>
<th>General Properties</th>
<th>Mechanical Support</th>
<th>Surface Properties</th>
</tr>
</thead>
<tbody>
<tr>
<td>small outer diameter</td>
<td>high plastic formability for a large expansion</td>
<td>low thrombogenicity</td>
</tr>
<tr>
<td>thin struts</td>
<td>high modulus of elasticity (Young’s modulus) to minimize the</td>
<td>high biocompatibility</td>
</tr>
<tr>
<td>high flexibility</td>
<td>recoil</td>
<td>non-corrodng</td>
</tr>
<tr>
<td>high expandability</td>
<td></td>
<td>growability of endothelial cells</td>
</tr>
<tr>
<td>low elastic recoil after expansion</td>
<td></td>
<td></td>
</tr>
<tr>
<td>low longitudinal contraction after expansion</td>
<td></td>
<td></td>
</tr>
<tr>
<td>sufficient radial stability</td>
<td></td>
<td></td>
</tr>
<tr>
<td>sufficient x-ray visibility</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Balloon expandable stents require materials with low yield stress, high elastic modulus and is work hardening through expansion for high strength [9]. They are manufactured in small diameters and dilated to the expanded shape when placed inside the vessel. The standard and most widely used material for balloon expandable stents is stainless steel 316L. This material has high corrosion resistance and low carbon content with molybdenum and niobium additions. Alternative materials for balloon-expandable stents include platinum alloys, niobium alloys and cobalt alloys. This is due to the following characteristics: better radiopacity, higher strength, high corrosion resistance, and biocompatibility [9].

Self-expanding stents are fabricated in the expanded shape. The stent is then compressed and constrained before deployment in the artery. When placing in the artery, the stent is self-expanded to the original diameter. The most widely used material is nickel-titanium alloy (nitinol) which can recover elastic deformations of up to 10% (large elastic strain and low elastic modulus) [9].

Metallic materials have been most preferable in stent manufacturing owing to radiopacity capability. In addition, they have the following benefits; (i) nearly ideal mechanical properties, (ii) availability of a fast laser-based machining process, (iii) well established subsequent treatments of the stent exist for steel and (iv) comparatively cheap material [71, 92].

Alloyed metal is found to give a better performance where their material properties suit the requirement of cardiovascular implants. Different alloying materials has a different performance [93]. The Co-Cr alloys are found stronger than stainless steel
Chapter 3: Laser cutting of coronary stents for cardiovascular treatments

316L and allow a thinner strut to be produced without affecting the radial strength. Furthermore, the Co-Cr alloys are denser than 316L and thus can provide excellent radiopacity. For precious metal, platinum-10% iridium alloys have been utilised in a stent application while a pure precious metal is mechanically weak to suit the stent material’s requirement. It was reported that precious metal (platinum, iridium and gold) are favourable materials to other alloys to further increase the radiopacity [94]. The potential of using plastics for stent materials (polymer) will be discussed in Section 3.10.

3.5 Stent fabrication methods

Stents can be manufactured from raw material shape such as sheet, wire, ribbon or tubing. Wire and tubing are the widely used profiles in producing expandable and self-expanding stents [9]. Conventional stent fabrication methods include photochemical etching, knitting, welding and braiding. Some examples of stents fabricated with traditional methods are shown in Figure 3.3.

![Figure 3.3: Example of stents fabricated with traditional methods. (a) Braided stent fabricated from cobalt alloy wire, (b) Knitted stent made of tantalum wire and (c) Welded stent made of nitinol wire [9].](image)

By far, the vast majority of coronary stents are produced by laser cutting from tubing [9]. This is the most effective method in processing stents as compared to other traditional methods that take considerable processing times [95]. Laser cutting processes are reliable in cutting large production volumes making this the most
preferable method in stent production. Laser cutting provides the following advantages in comparison with conventional methods: high precision, fast and low cost. The materials suitable for laser micro-cutting in stent manufacture include stainless steel 316L [11, 96-98], stainless steel 316 LVM [99], nickel-titanium alloy [100] and cobalt chromium [101]. Figure 3.4 shows an example of stent fabricated with a laser. Table 3.3 lists the physical and mechanical properties of selected materials, stent geometry and fabrication methods [102]. The advantages and disadvantages of different fabrication methods are listed in Table 3.4.

Figure 3.4: Example of coronary stent cut by laser [8].
Table 3.3: A Summary of materials used in balloon-expandable and self-expandable stents, different stent forms, stent fabrication, stent geometry and additions [102].

<table>
<thead>
<tr>
<th>Materials</th>
<th>Balloon expandable stents</th>
<th>Self-expanding stents</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Stainless steel (vast majority)</td>
<td>Nitinol (vast majority)</td>
</tr>
<tr>
<td></td>
<td>Tantalum</td>
<td>Cobalt alloy</td>
</tr>
<tr>
<td></td>
<td>Martensitic nitinol</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Platinum iridium</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Polymers</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Niobium alloy</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Cobalt alloy</td>
<td></td>
</tr>
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</table>

<table>
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<tr>
<th>Form</th>
<th>Wire</th>
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<tbody>
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<td>Bridge, S7, S660,(stainless steel, welded rings)</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Angiostent (platinum iridium)</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Strecker (tantalum)</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Expander (nitinol)</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Vast majority</td>
<td></td>
</tr>
</tbody>
</table>

<table>
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<tr>
<th>Form</th>
<th>Sheet</th>
<th></th>
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<tbody>
<tr>
<td></td>
<td>NIR (stainless steel)</td>
<td></td>
</tr>
<tr>
<td></td>
<td>ZR1 (stainless steel)</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Endotax (nitinol)</td>
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<table>
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<tr>
<th>Form</th>
<th>Ribbon</th>
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<tbody>
<tr>
<td></td>
<td>Horizon prostatic (nitinol)</td>
<td></td>
</tr>
<tr>
<td></td>
<td>EndoCoil, esophacoil (nitinol)</td>
<td></td>
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</tbody>
</table>

<table>
<thead>
<tr>
<th>Fabrication</th>
<th>Laser cutting</th>
<th>Vast majority</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Photochemical etching</td>
<td>NIR</td>
</tr>
<tr>
<td></td>
<td>Braiding</td>
<td>Nitinol sheet</td>
</tr>
<tr>
<td></td>
<td>Knitting</td>
<td>Wallstent (cobalt alloy)</td>
</tr>
<tr>
<td></td>
<td>Water jet</td>
<td>Strecker (tantalum)</td>
</tr>
<tr>
<td></td>
<td></td>
<td>SCS, SCS-Z stent</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Geometry</th>
<th>Helical spring</th>
<th>Periodic peak to peak connection</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>No/minimal connections</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Woven</td>
<td>Braided</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Knitted</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Geometry</th>
<th>Sequential rings</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Open cells</td>
<td>Peak-peak connections</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Peak-valley connections</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Midstruts connections</td>
</tr>
<tr>
<td></td>
<td>Closed cells</td>
<td>Regular peak-peak connection</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Non-flex connector</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Flex connector</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Combined connector</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Additions</th>
<th>Covering</th>
<th>WallGraft: coiled nitinol framework, ePTFE covering</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Radiopaque coating</td>
<td>Gold, silicone carbide over gold</td>
</tr>
<tr>
<td></td>
<td>Biocompatibility coatings</td>
<td>Tantalum coatings, phosphorylcholine, carbon coating</td>
</tr>
<tr>
<td></td>
<td>Drug eluting coating</td>
<td>silicone carbide</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Rapamicyne, paclitaxel</td>
</tr>
</tbody>
</table>
Table 3.4: The advantages and disadvantages of different fabrication methods [9, 103, 104].

<table>
<thead>
<tr>
<th>FABRICATION</th>
<th>ADVANTAGES</th>
<th>DISADVANTAGES</th>
</tr>
</thead>
<tbody>
<tr>
<td>LASER CUTTING</td>
<td>Intricate patterns, able to process small diameter tube, reliable, fast, high precision and low cost</td>
<td>Heat effects</td>
</tr>
<tr>
<td>WELDING</td>
<td>Adjustable flexibility and a wide range of size and length</td>
<td>Radial strength is usually less than the ones processed with laser</td>
</tr>
<tr>
<td>BRAIDING, KNITTING</td>
<td>Simple process, adjustable flexibility</td>
<td>Slow process</td>
</tr>
<tr>
<td>PHOTOCHEMICAL ETCHING</td>
<td>Large numbers of parts can be processed in a single run, complex patterns can be produced</td>
<td>Require extremely clean operating conditions</td>
</tr>
</tbody>
</table>

3.6 Laser Cutting of Coronary Stents

3.6.1 Laser stent cutting – basic process description

In general, a stent-cutting machine requires one rotation and 2 linear axes. Usually, prior to processing, computer aided design (CAD) data of the stent with the desired configuration is generated [71, 96], which is further analysed with finite element method for its optimum performance, by considering the material characteristics and the mechanical performance (elasticity, ductility, material strength) [9, 71]. Example of a stent design is shown in Figure 3.5 [71]. Finally the data are fed to the laser processing system and then cutting of the desired pattern can be performed with the computer numerical control (CNC) programming. Figure 3.6 shows the schematic drawing of a laser based stent cutter, where the laser beam is focused by a lens and a process gas is introduced coaxially with the laser beam to assist the material removal [71].

![Optimised stent design](image)
Chapter 3: Laser cutting of coronary stents for cardiovascular treatments

Figure 3.6: Schematically drawn setup of a laser-based stent cutter [71].

For metals, the cutting process can be performed by melting and gas blowing, by exothermic reaction with the supplied oxygen gas and by vaporisation or sublimation [1, 57]. The vaporisation cutting can only be applied by using a short pulsed laser with a high peak power [74]. The advantage of sublimation is that melting and heat affected zones can be minimised or even avoided.

Type of laser utilised in stent cutting are listed as follows: i) Flash pumped Nd:YAG laser, ii) Fibre laser, iii) Disk laser, iv) Nanosecond laser, v) Picosecond laser, and vi) Femtosecond laser. The lasers with longer pulses have longer beam/material interaction time which resulted in heat affected zone to the stent workpiece. The shorter pulse durations (femtosecond lasers) result in no or minimum heat affected zone but, slower speed. Beam quality is also an important key in stent processing. Beam quality, $M^2 = 1$ is preferable and provides superior beam quality where desired spot size and laser power density can be achieved.
3.6.2 Important quality aspects in laser stent cutting

In laser cutting processes, the material removal mechanism involves localised heat application by the focused laser beam which turns the target material into liquid and gas. This will set a challenge in the fabrication of such an intricate design in the miniature form such as stents. The important quality aspects that should be considered for high quality laser stent fabrication are listed as follows.

(i) **Narrow and consistent kerf width** is required to allow some space for electropolishing, which involves material removal that causes strut width reduction. It is important to maintain the cut width because the intricacy of cut path is also required for uniform stent expansion. Typical kerf widths of stents cut by laser is around 20 – 30 µm [71, 97, 98].

(ii) **Absence of heat affected zone (HAZ)** on cut samples. Excessive heat from laser beam causes thermal effect to the stent material resulting in embrittlement and undesirable heat affected zones (where the material microstructure has changed) adjacent to the laser cut. This deficiency can cause stent rupture when the stent is expanding during implantation [8]. Typical HAZ width in laser stent cutting is 30 µm. The use of nitrogen as an assist gas instead of oxygen can reduce heat affected zone due to weak oxidation effect. Post processing procedures are normally required to remove the HAZ (see Section 3.9). HAZ can be minimised or eliminated with the use of ultra-short pulse (picosecond and femtosecond pulse width) lasers.

(iii) **Smooth cut edges** - Stents require exceptionally smooth surface since any imperfect edge may cause irritation during or after implanted in human body [105]. Striations on the cut edge must be minimum [98]. Stent suitable for implantation within a blood vessel require an surface finish, $R_a$ of about 0.2 down to 0.05 µm [106]. Typical surface finishes, $R_a$ in laser cutting is around 0.4 to 1 µm depending on the processing conditions. The surface finish can be
improved by utilising a high pulse repetition rate in laser cutting which provides high pulse overlapping and thus gives a finer cut surface. Post processing procedures are normally required to improve the surface finishes.

(iv) **High geometry accuracy, clean edges with no dross or recast layers** – dross and recast layer need to be removed for high stent accuracy and biocompatibility. Dross can be minimised by applying a high pressure inert gas during the process. Furthermore, underwater machining is also useful in reducing dross and recast.

(v) **Absence of back-wall damage** – backwall damages may occur in the laser cutting process by the laser beam with molten particles adhering to the opposite side of the wall. These damages have to be cleaned or avoided.

Production of human coronary artery stents demand high precision and high quality surface finishes [107]. Currently laser cutting always require post processing to produce high quality stents. The surface quality of stent plays a major role since it has a significant influence on biocompatibility. Post-processing procedures involved in stents manufacturing include electropolishing, acid pickling and soft etching [8]. Details of post processing are described in Section 3.9.

### 3.6.3 Process parameters affecting laser stent cutting

Laser stent cutting processes are affected by various processing parameters which include peak pulse power, repetition rate, pulse duration, cutting speed and assist gas pressure and type. The parameter selection must be clearly understood for different lasers and different materials since it significantly affect the final cutting results in terms of geometric and metallurgical quality [18]. This section discusses the impacts of the most important laser processing parameters on the geometric and metallurgical
quality of the cuts. A detailed explanation of the main cutting parameters is attempted in the following sections.

3.6.3.1 Laser peak pulse power

Laser peak pulse power has a significant effect on the material removal mechanism. As the peak pulse power (in Watt) increases, more intensive energy is transferred into the workpiece within a short period of time. More melting and material evaporation and even plasma generation will take place. Excessive laser energy input has been observed to result in wider kerf widths, a wider HAZ and excessive melting which may effect the surface roughness [108].

The peak pulse power can be expressed as [71, 109]:

\[ P_p = \frac{E}{t_p} \]  \hspace{1cm} (3.1)

where \( P_p \) is the peak pulse power, \( E \) is the pulse energy and \( t_p \) is the pulse duration. Peak pulse power can be controlled by adjusting the pulse energy and pulse duration as these parameters are interdependent. For certain lasers, the pulse width is fixed and not user adjustable.

Kleine et al. [108] showed the relationship between laser peak pulse power and the average roughness in fibre laser cutting of stainless steel. The average roughness was found to increase with increasing peak pulse power due to the striations effect resulting from the hydrodynamic instability during the melting process (see Figure 3.7). Ahn et al. [98] also found that laser peak power plays a major role in affecting the kerf width. Meszlenyi et al. [97] reported that increasing power causes increased heat effects in the cut samples.
Figure 3.7: Average surface roughness as a function of peak pulse power in fibre laser cutting [108].

3.6.3.2 Pulse repetition rate

Laser pulse repetition rate or laser pulse frequency influenced the cut surface quality. Kleine et al., [108] studied the effects of repetition rate on the laser cut surface and found that surface roughness decreases with increasing repetition rate as shown in Figure 3.8 due to a high pulse-to-pulse overlap as the repetition rate increases [11].

Figure 3.8: Surface roughness, $R_a$ as a function of laser pulse frequency during fibre laser cutting of stainless steel [11].
The pulse overlap (PO) can be obtained with the following expression:

$$PO = 1 - \frac{v_s}{d_{effective} \cdot f}$$  \hspace{1cm} (3.2)$$

where $v_s$ is the cutting speed, $d_{effective}$ is the focus beam diameter and $f$ is the pulse repetition rate.

Figure 3.9 illustrates the schematic of the pulse overlapping. A high repetition rate significantly increases the pulse overlapping leading to a more effective material removal as shown in Figure 3.10. Shallow material removal is obtained at low pulse overlapping (PO = 14%) as compared to the one with high pulse overlap (PO = 99%) [110].

Figure 3.9: Schematic diagram of the pulse overlap [110].

Figure 3.10: SEM images to demonstrate the comparison of two PO values after one scan, (a) PO=99% and (b) PO=14% [110].
3.6.3.3 Cutting speed

Cutting speed has been found to have a significant influence on the kerf width and surface quality. Increasing the cutting speed leads to a reduction of surface roughness after a critical cutting speed is achieved. Less interaction time between the laser beam and material reduced the energy supplied to the cut kerf and thus less amount of material is melted. The available energy within the interaction zone decreases, hence reduce the kerf width as shown in Figure 3.11 [39].

![Figure 3.11: Influence of the cutting speed on the kerf width [39].](image)

3.6.3.4 Pulse duration (or pulse length)

Pulse duration or pulse length or pulse width is an important parameter in laser cutting because it controls the power density delivered into the workpiece. In today’s laser technology, the pulse duration is available from milliseconds down to a few femtosecond ($10^{-13} - 10^{-15}$ s) for the laser based stent cutting processes. Pulse duration affects the beam and material interaction time. Pulse duration is one of the most significant parameters related to the HAZ formation. For example, at femtosecond pulse duration the HAZ formation is almost absent. Ultra-short pulses were also suggested to potentially reduce recast layer [11].
The relationship between pulse duration and the formation of HAZ can be explained under three different cases (time scales). To describe the process, the following notations are used: $T_e$ is electron cooling time (in order of 1 ps), $T_i$ is the lattice heating time and $T_l$ is the laser pulse duration. It can be deduced that $T_e << T_i$ due to the electron heat capacity being lower than lattice heat capacity.

First when $T_l$ is in the millisecond regime ($T_l > 1 \text{ ms} >> T_i >> T_e$), the beam material interaction time is far longer than the electron cooling time. This allows adequate time for electron relaxation which subsequently transfers the energy to the lattice. The material in the vicinity of the active laser beam receives high amount of energy which transforms it into heat, resulting in material melting (dominant mechanism) and vaporisation. In this ms timeframe, the main material removal mechanism is based on melting and an assist gas jet is required to remove the molten metal.

Second, when $T_l$ is in the nanosecond regime ($T_l > 1 \text{ ns} >> T_i >> T_e$), the electrons of the material being cut absorb the laser energy and they still have adequate time to transfer this energy to the lattice. In the ns timeframe, the material is melted to the liquid state and it also reaches the vapour state and capable to produce plasma depending on the beam intensity level. A smaller HAZ can be generated in this regime compared to the millisecond/continuous wave regime.

Third, when $T_l$ is in the femtosecond regime ($T_l << T_e << T_i$), the pulse duration is shorter than the electron cooling time. The electrons absorb the high energy (always the case for ultra-short pulse laser) and transfer this energy to the lattice ions (in about 1 ps time scale). The ions receive the high laser energy which is sufficient to break the lattice bonding. A direct solid-vapour transition could take place and heat conduction can be neglected due to insufficient time to transfer energy to the neighbouring ions; therefore, HAZ formation is negligible. This regime is suitable for high quality stent cutting applications [38, 74, 111-113]. However, due to low average power of the fs lasers, the cutting speed is generally much lower compared to continuous wave or longer pulse lasers. The advantages of stent cutting with ultra-short pulse lasers will be discussed in details in Section 3.7.
3.6.3.5 Assist gas

One very important factor to consider in laser cutting is choosing the right assist gas. Most common gases used in laser cutting are oxygen and nitrogen. In addition to the function of lens protection, the main aim use an assist gas is to eject the laser melted material, leading to less dross and recast in stent machining. An appropriate assist gas pressure is required in order to reduce dross level. Oxygen is widely used as an assist gas in stent cutting to gain additional thermal energy through exothermic reactions with the metallic materials at high temperatures. However oxygen oxidises the cut surface and the additional heat generated can distorted the workpiece material. Nitrogen minimises these effects because it makes the process inert. Nitrogen is preferable in cutting stainless steel 316L for clean edges. However, higher pressure is required for nitrogen to blow away the molten material [114]. Argon is preferable in cutting titanium or nickel titanium because the incompatibility of these metals with other assists gas.

Meszlenyi et al. [97], Kleine et al. [11], Sudheer et al. [115] and Meng et al. [96] suggested the use of nitrogen as an assists gas instead of oxygen, which can reduce heat affected zone due to weak oxidation effect. Kleine et al. [11] also found that high assist gas pressure (greater than 7 bar) potentially reduces the recast layer thickness. Zheng et al. [116] performed studies on femtosecond laser processing of nitinol sheets and used a stream of nitrogen gas to reduce debris.

3.7 Progress and Development in Laser Based Stent Cutting Technology

The first laser cut stent (Palmaz-Schatz stent) was approved by Food and Drug Association for clinical use in the United States in 1994 [8]. To date, most of the commercial coronary stents are manufactured by laser cutting of tubular materials.
Flash lamp pumped Nd:YAG lasers are the primary laser source used in commercial stent manufacturing in the beginning of application.

Early work of Nd:YAG laser micromachining of stent includes the work by Kathuria in 1998 [10]. This was followed by further research by Kathuria [92, 117, 118], Raval et. al [119] and Meszlenyi et. al [97]. A focus diameter of 20-50 µm was used based this type of lasers [120]. The typical cutting parameters by using Nd:YAG lasers are summarised in Table 3.5.

Table 3.5: Typical cutting parameters by using Nd:YAG lasers summarised from following [8, 10, 97, 115, 119].

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Range</th>
</tr>
</thead>
<tbody>
<tr>
<td>Average power</td>
<td>2 W to 20 W</td>
</tr>
<tr>
<td>Pulse width</td>
<td>2 µs - 0.16 ms</td>
</tr>
<tr>
<td>Frequency</td>
<td>30 Hz to 10 kHz</td>
</tr>
<tr>
<td>Cutting speed</td>
<td>0.2 mm/s - 4 mm/s</td>
</tr>
</tbody>
</table>

Conventional Nd:YAG lasers operated within the microsecond and milisecond pulse range has limitations due to the fact that the material removal mechanism is based on melting by the laser beam and blown away from the cut zone with an assist gas due to long beam and material interaction time. This long pulse machining resulted in HAZ generation at the vicinity of the cut regions. The melting also results in significant dross, slag, recast and oxide layer, and backwall damage which need to be post-processed. Figure 3.12 demonstrates the quality issue showing the oxide layer and slag attached to the inner surface of tube associated with Nd:YAG laser cutting [121]. Figure 3.13 shows the thermal effect on the opposite site of the tube [97].
Figure 3.12: Nd:YAG laser cut stents (a) cutting zone, (b) outer surface and (c) inner surface, covered with oxide layer and slag [121].

Figure 3.13: Higher average power cause heat effect to the opposite site of the tube[97].

Related research [71, 92, 95, 115, 117, 118] show that Nd:YAG lasers are one of the established tools for micro cutting applications. Explosive growth of laser technology over the last decade led to the development of diode pumped Nd:YAG lasers, fibre lasers and disc lasers competing against the commonly applied flash lamp pumped Nd:YAG laser due to the following advantages of the new type of solid state lasers; i) higher power stability, ii) better energy efficiency from the previous 1-5% to 20-40% in converting the electrical input energy to optical energy, iii) eliminates flash lamp pump changes, iv) low maintenance and v) smaller footprint [11, 96].
Drawbacks such as low energy efficiency, large foot print, low density of focus power, large heat affected zone are the downfalls why Nd:YAG lasers are struggling in competing with the new types of lasers [71].

The first micro cutting by using a fibre laser in the stent applications was reported by Kleine et al. [11]. They claimed that fibre laser was more reliable due to excellent beam quality, smaller kerf widths which were suitable for micro machining. The measured spot size was in the range between 18 to 20 µm. Liu et al. [120] proposed the typical fibre laser cutting parameters used in micromachining by referring to the work of Kleine et al. [11] as shown in Table 3.6. This table also includes the parameters used by Meng et al [96].

<table>
<thead>
<tr>
<th>Average power</th>
<th>20 W - 50 W</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pulse width</td>
<td>0.05 ms - 1 ms</td>
</tr>
<tr>
<td>Frequency</td>
<td>0.5 kHz - 5 kHz</td>
</tr>
<tr>
<td>Cutting speed</td>
<td>5 mm/s - 15 mm/s</td>
</tr>
</tbody>
</table>

Cutting with a fibre laser is not free from quality issue due to the long pulse duration and melting. For quality improvement, Liu et al. [120] attempted to design a fibre laser to be pulse length, pulse repetition and laser power tunable in order to achieve a good quality cut and reduce post processing requirements. Work by Meng et al. [96] demonstrated a high quality metallic cardiovascular stent cutting system based on a pulsed fibre laser by implementing the method introduced by Liu et al. [120].

Stent cutting by means of Nd:YAG and fibre lasers, producing pulses with durations in the range of microseconds \((10^6 \text{ s})\) to millisecond \((10^3 \text{ s})\) has some limitations. The melting mechanism at this pulse duration again leads to a significant burr formation at the cutting edges and deposition of solidified droplets, sticking on the tube surface. In long pulsed laser machining of metals, it is well known that the energy shielding effect of the plasma plume formed by ejected materials from the ablation crater
greatly lowers the efficiency of laser energy coupling to the work piece. Furthermore, machining quality suffers from this plasma plume that causes prolonged heating and melting effects [122].

As the medical device industry has advanced, it has become extremely important to improve time to market, throughput, and manufacturing tolerances. In the area of medical device, thermal management is the main challenge especially when the cutting is performed using long pulsed duration lasers. The advancement of the laser industry has driven the development of shorter pulse lasers including nano, pico- and femtosecond pulse duration range which would offer new opportunities to minimise and eliminate HAZ [46, 116, 123-130]. The material removal in short and ultra-short pulse laser machining is based on the ablation which heat the material up to the vaporisation point. They further reduce the heat propagation into the material by reducing the beam and material interaction time. Furthermore, picosecond and femtosecond laser pulses have a much lower heat load compared to pulsed laser radiation with nano- and microsecond pulse lengths [123]. This characteristic is desirable in cutting sensitive and delicate materials to satisfy the stringent requirement in stent manufacturing. The use of ultra-short pulse lasers (picosecond and femtosecond pulses) is a promising alternative to the longer pulse lasers for machining thin and intricate materials for coronary stent applications. The advantages of the ultra-short pulse lasers were summarised by Meijer et al. [74].

The investigations on ps and fs laser cutting of coronary stents are relatively new. Picosecond machining of metallic sheets were discussed and investigated by several groups [123, 124, 130]. Jandeleit et al. [123] in their work found the removal rate per pulse of nano- and picosecond laser pulses allow the precise removal of material. For cutting tubing material for coronary stent applications, no specific study has been found reported at the start of this PhD in 2009.

Femtosecond laser cutting of thin nitinol sheet was performed by Li et al. [100] at a 150 fs pulse duration. They concluded that femtosecond laser cutting was still based on a thermal mechanism. High ablation rate cutting with high laser fluence causes
significant recast and debris deposited around the perimeter of the ablation region as shown in Figure 3.14.

![Figure 3.14: Prominent recast is found around the channel cut with femtosecond laser fluence as high as 40 J/cm² [100].](image)

The debris formation deteriorates the feature quality and functionality of the components. It also reduces ablation efficiency since debris from the previous scan blocks the laser beam path for the next scan. Femtosecond laser machining of nitinol produced less heat load. However, the recast and debris were still present around the perimeter of the cut edges [100].

Over the last decade, ultra-short pulse (picosecond to femtosecond) lasers have been available for industrial applications, especially for high precision processing. The use of an ultra-short pulse laser is a promising alternative to the longer pulse lasers for machining thin and intricate materials for coronary stent applications. The use of femtosecond laser machining has been shown to reduce the size of HAZs, due to short beam/material interaction time. However, the main obstacle in applying this technology is the debris and recast formation, where the surface of the ablated material is usually surrounded by re-deposition of vaporised material that has to be subsequently removed by other methods. In addition, the cutting speed is very low.
3.8 Approaches in improving the laser stent cutting process

In order to obtain high quality machining results, one should not only focus on choosing the right laser with the correct processing parameters. Additional techniques or approaches have to be considered when dealing with debris, recast, burrs and heat affected zones in laser cutting. Examples of these techniques for removing defects during the laser cutting process include the introduction of water assisted processes and other protection methods.

3.8.1 Water assisted laser cutting of stents

One technique used for improving cut quality is water assisted laser cutting. This method is by either introducing a water flow or submerged water during the cutting process. Meng et al. [96] designed and utilised the tube cooling equipment that could pump water through the tubes during the process. They found that this technique managed to reduce the heat affected zone and prevented damages to the opposite surface of the tube in fibre laser cutting of stainless steels. A patent by Merdan and Shedlov [131] discloses a backwall damage prevention method in stent manufacturing. A conduit fluid source through the tube was proposed to remove the debris and assist in removing heat that is generated during the cutting. One of the suggested fluids to be used is water. Another possible method is the use of the water-jet guided laser technology introduced by Synova. With this method, a water jet was coupled with a laser beam during the process in order to reduce the heat affected zone and improving surface quality [132].

3.8.2 Other methods

Preventing back wall damage is a challenge in a cutting such small and thin tubes in coronary stent manufacture. The ejected materials from a cut kerf in a form of hot particles adhere and create backwall damages to the opposite wall of the tube. On the
other hand, the laser beam transmission was also not prevented from reaching the opposite wall causing deterioration to the backwall. Kathuria [118] and Raval et al. [119] managed to protect the laser beam energy transmitting to the backwall by inserting Teflon and Teflon coated brass through the inner wall.

3.9 Stent finishing: post-processing stages in laser stent manufacturing

Laser cut stents are not the final product for industrial coronary stents. They must be free from any dross and imperfections produced from the cutting process. The component must be smoothed (edges and cut surface), passivated for corrosion resistance and must be cleaned and also sterilised. Figure 3.15 shows a process flow in producing stents including a series of post processing steps.

Figure 3.15: Post processing stages after laser cutting process [133] [121] [134] [107].

Zhou et al. [133], Raval et al. [121], Kathuria [134] and Geller et al. [107] applied acid pickling as a first post-processing step to remove dross, oxide layers and slags. Pickling technique is performed by immersing the stent in a tank containing dilute HCl acid contains the following mixture (HCl = 7%, FeCl₂ = 8% and H₂O balance [134]. The other method to replace acid pickling is by applying soft etching. This
involves electrochemical etching with a weak acid. Samples can also be treated by ultrasonic cleaning with the following solution; ‘Susbreito’ (contains 9.4% of HNO$_3$), and a surface active agent (contains a small proportion of chelate acid and balance of 90% H$_2$O) [134].

After acid pickling, annealing process is employed to make the stents implantable (sufficiently soft and ductile). The purpose of annealing is to treat the stents to make it flexible and expandable during the implantation process. In the annealing process, the stents were heated in vacuum from room temperature to a temperature of 1000$^\circ$C and then were kept in the chosen temperatures respectively for 1 hour duration. The samples were then slowly cooled in vacuum to room temperature [133].

Electropolishing was the treatment taken after the samples have undergone acid pickling and annealing processes to obtain a sufficiently smooth surface of stents [105]. Eletropolishing is the process of enhancing and smoothing the surface of stents in an electrolyte solution containing H$_2$SO$_4$ (95-97%), H$_3$PO$_4$ (85 wt%) and H$_2$O (balance) [99]. Optimal parameters must be used during the process to obtain a high surface quality and for a minimum weight loss of stents [107, 121]. Figure 3.16 shows the stent strut surface before and after undergo multiple post processing. Figure 3.17 shows the quality of stent surface after completing the electropolishing process.

![Figure 3.16: Stent strut (a) as cut surface (b) acid pickled (c) annealed (d) electropolished [133].](image)
After the electropolishing, passivation process is undertaken to improve stents’ surface characteristics to enhance the corrosion resistance while placed in biological environment [99, 135]. Passivation is carried out by dipping the stent materials in a solution containing 25% (v/v) concentrated HNO₃ and deionised water (balance) for 30 minutes [135]. Nitinol stent can be passivated by using nitric acid solution. The passivation treatment reduces the content of Ni or NiO and increases the surface with TiO₂ content which improves the corrosion behaviour of the materials. Passivation treatment by using acid nitric solution (oxidising substances) is the typical passivation method for certain materials such as stainless steels, cobalt-chromium alloys and for titanium alloys [136].

If high quality stents in laser cutting can be achieved, acid pickling process can be eliminated or time taken for the process could be minimised. On the other hand, electropolishing time also could also be reduced. Annealing process could not be eliminated since this heat treatment process is essential to making the stent ductile.
and implantable. Passivation treatment is important to improve the corrosion resistance of the materials. Most of the post-processing procedures involve the use of chemicals and involving hazardous processes. On the other hand post processing is a high cost, time consuming and labour intensive process and also de-scales the stent size. In Geller et al.’s [107] work, they explained that acid pickling and electropolishing reduce the stent strut width between 28 to 34.7 % of the original strut width. Therefore, the optimal method for post processing should be coupled with laser cutting procedures.

3.10 Challenges and Future Prospects

Due to the stringent requirement of the stent devices, challenges in stent cutting include: i) to maintain narrow and consistent kerf width and high geometry accuracy, ii) absence of HAZ, iii) smooth and high quality cut edges, iv) clean edge with no dross or recast layer, and v) absence of backwall damage. In the early development of laser stent cutting, the processes struggled with the cut quality issues. Manufacturing of stents have to focus on the stability and uniformity of cutting process. With the growth of laser technology, the emergence of ultra-short pulse lasers provides stable process and uniformity in the product quality. The ultra short pulse laser cutting also improves the surface finish, tolerance and higher degree of process control which relates to cost improvement. However, the disadvantage of ultra short pulse laser processing is the slow cutting speed as a result of low average powers of these lasers currently available. Further research development is therefore necessary for improving speed to meet the industrial yield. Thus, proper customisation between laser type, laser characteristics and workpiece specifications is required to meet stent quality requirement. Furthermore, new materials for stent materials (metal alloys), polymer and magnesium bio-absorbable stent requires new studies to understand how the material change impact on the laser cutting process. Modelling and simulation together with in-process monitoring and control will
further enhance the understanding of the process and improvement in quality and productivity.

Polymer stents (future stents) are more difficult to cut with long pulses laser. This is due to the low melting points of polymer material which make the control of thermal damage even more difficult. The athermal ablation capability of ultrashort pulse lasers with an UV wavelength plays an important role in the manufacturing of these materials. Ultra-short pulse laser machining can be used to cut different kinds of materials with minimum mechanical and thermal damages. This novel machining technique is presently under investigation, and its potential for the generation of special and new types of stents is to be explored [71]. Kathuria [134] and Knowles et al. [137] suggested that polymer materials could be cut at the appropriate laser wavelength in the UV region. Bauer et al. [138] illustrates the HAZ formation in cutting polymers with CO\textsubscript{2} and femtosecond lasers and undetectable HAZ is possible with the femtosecond pulse laser as shown in Figure 3.18. Figure 3.19 show examples of polymer stents fabricated with femtosecond lasers.

![Figure 3.18: Diagram of heat-affected zones in biodegradable polymers of CO\textsubscript{2} and fs laser machined cuts [138].](image)

Figure 3.18: Diagram of heat-affected zones in biodegradable polymers of CO\textsubscript{2} and fs laser machined cuts [138].
Chen [8] proposed that the ideal stent cutting system should comprise the following features as shown in Figure 3.20 for a high accuracy stent processing towards cold machining.

Figure 3.19: A medical stent micro machined from a biodegradable polymer using a femtosecond laser (Courtesy of LZH/ Cortronik) [139].

Figure 3.20: Features required for an ideal stent cutting system [8].
3.11 Summary

Considerable experiences and research have been gathered in the field of laser processing for medical stents fabrication. Various laser applications in laser stent cutting have been reviewed. The ultra-short pulse lasers are gaining greater attention in the future stents manufacturing. Laser stent cutting requires system optimisation for high yield, high quality and low cost. From the extensive reviews, the current knowledge gaps have been determined for further research. It was found that fibre laser water assisted cutting has been undertaken however there is no scientific study available in explaining the performance of fibre laser cutting with and without water flow. On the other hand, there is also gap in the research where there is only a few research reported regarding the picosecond laser, however not specifically in cutting tubular materials for stent applications. Femtosecond laser cutting gives a precise and high quality cut however producing debris vicinity to the cut area which requires further research for quality improvement. The presented works in the following experimental chapters (5, 6 and 7) were carried out to fill the aforementioned research gaps in the area of laser stent cutting.
Chapter 4

Experimental Apparatus and Analytical Equipment

4.1 Introduction

This chapter introduces the experimental apparatus and analytical equipment that have been utilised in this research work. Three different laser sources with different range of pulse durations have been used. Fibre laser and picosecond laser was integrated with the Swisstec Micro T15 platform while the femtosecond laser system was based on the optical train. The specifications for each laser source, the stent cutting platform and the analytical equipment for samples’ characterisation is described in the following section. The experimental methodology is explained in detail in each related experimental chapter.
4.2 Stent cutting system – Micro T15

The machining system used in part of this work is the compact and high performance tube cutting system Micro T15 optimised for stent production as shown in Figure 4.1. This system includes a CNC motion control (3 axes: rotation, transverse and height), a beam collimator, a cutting head with a focusing lens, a coaxial gas nozzle and a charged coupled device (CCD) vision system looking directly down to the nozzle at the cutting area. The coaxial assist gas nozzle had an exit diameter of 0.5 mm. The nozzle was optimised for a maximum velocity and coaxial gas flow. This machine has the capability to pump the water inside the tube as an additional cooling option during the cutting process. The laser beam remains stationary and the tube rotates and traverses automatically during the laser machining process.

Figure 4.1: Swisstec Micro T15 machining system.

In this work, the Micro T15 machining system used was integrated with two different types of lasers: i) Micro T15 integrated with fibre laser system, ii) Micro T15 integrated with picosecond laser system. The technical specifications of the fibre
laser and picosecond laser that have been used are described in the following subheadings.

4.2.1 Fibre laser

A GSI JK100 FL fibre laser system with the following characteristics was employed in this research work:

- Wavelength, $\lambda = 1080 \pm 5$ nm
- Beam quality, $M^2 < 1.1$
- Maximum average power = 100 Watt
- Maximum modulation frequency = 50 kHz
- Laser mode: TEM$_{00}$

This laser source has a Gaussian intensity distribution and nearly Gaussian pulse shape. It has been used to cut a simple structure of stainless steel 316L in dry and wet cutting conditions (in millisecond pulse duration) for a cut quality comparison. Chapter 5 details the experimental work performed with the fibre laser.

4.2.2 Picosecond laser

The picosecond laser used in this research was picosecond laser Trumpf TruMicro 5350 with the following characteristics:

- Wavelength, $\lambda = 343$ nm (UV)
- Beam quality, $M^2 = 1.2$
- Maximum average power = 10 Watt
- Maximum modulation frequency = 800 kHz
This laser also has a Gaussian intensity distribution and a nearly Gaussian pulse shape. It has been chosen to operate in the UV wavelength range to study the advantages of cutting metal and polymer materials at this wavelength and at 6 ps pulse duration. Chapter 6 reports the details of the experimental work carried out with the picosecond laser.

### 4.3 Femtosecond laser system

The femtosecond laser system employed in this research is based on the optical train as shown in Figure 4.2. This femtosecond laser is a Coherent Libra Ti: Sapphire System with the characteristics enlisted below:

- Wavelength, $\lambda = 800$ nm
- Maximum average power = 1 Watt
- Modulation frequency = 1 kHz
- Pulse width = 100 fs
- Original beam diameter = 6 mm

The laser beam was manipulated by computer controlled galvo scanner and it can be focused down to 100 $\mu$m spot size. This laser was used in the experimental work to study the cutting quality of the investigated materials with femtosecond pulses. At this range of pulse duration, high quality cuts were expected with no presence of HAZ. Details of the cutting process and results with femtosecond laser are reported in Chapter 7.
Figure 4.2: The optical setup for the Libra-S system (the red line denotes the laser beam path).
4.4 Analytical equipment

The laser processed samples were characterised by using different types of analytical equipment to examine the quality of the cuts. The analytical equipments used in this research are as follows:

4.4.1 Optical microscope

The cut samples were analysed via optical microscope to obtain measurements of the cut kerf width, HAZ and for samples imaging. The optical microscope used was Polyvar MET microscope (Figure 4.3) with adjustable magnifications from 2X up to 100X. It is coupled with CCD camera for automatic image transference to an integrated i-Solution DT image analysis software. This software allows the user to conduct multiple measurements and manipulation of images.

Figure 4.3: PolyVar-MET microscope.
4.4.2 Scanning electron microscope (SEM)

A filament based scanning electron microscope (SEM) by Hitachi High Technologies (S-3400N) (Figure 4.4) was used for high resolution imaging. It is equipped with secondary electron (SE) and back scattered electron (BSE) detectors and with magnification up to 300,000X and accelerating voltage up to 30 kV.

The SEM works on the principle of exposing the sample with a beam of electrons. The specimens must be conductive to allow the interaction with the electron and generate response from the specimens revealing the topography and other related information.

![Hitachi (S-3400N) scanning electron microscope.](image)

Figure 4.4: Hitachi (S-3400N) scanning electron microscope.

4.4.3 White light optical interferometer

4.4.3.1 Wyko NT-1100

Wyko NT-1100 white light interferometry (Figure 4.5) was utilised for assessing the characteristics of the cut surface. It is a fast and reliable non-contact measuring technique allowing a comprehensive analysis of 3D surfaces measured within the
range of sub-nanometer to millimeter roughness (0.1 nm to 1mm) [140]. The system is equipped with Wyko Vision analysis software for further visualisation and analysis options. The Wyko working principle is based on a splitted light beam which is reflected from the reference mirror and the specimen. The interference data from the reflected light beams produce signals containing the surface topography information.

![VEECO Wyko NT-1100 Interferometer](image)

**Figure 4.5: VEECO Wyko NT-1100 Interferometer.**

### 4.4.3.2 MicroXam Interferometer

MicroXam white light interferometry was also used to examine the cut surface characteristics (Figure 4.6). In this interferometer, a tungsten halogen lamp provides the light source. The working principle is similar to Wyko, i.e. it is based on splitted light beams which are reflected from both the reference mirror and the specimen. Then the interference pattern is then imaged to the CCD array, and subsequently read by a computer for its analysis.
4.4.4 MTS Nano Indenter XP

MTS Nano Indenter was used for hardness test in the vicinity of the cut area to determine the heat affected zone (Figure 4.7). The Nano Indenter XP is equipped with a motorised sample manipulation table with joystick control, optical imaging system and diamond indentation tip. It is connected to a personal computer with an analysis software package. The specifications of the MTS Nano Indenter XP system are enlisted below:

- Displacement resolution < 0.01 nm
- Load resolution = 50 nN (5.1 μgm)
- Maximum indentation depth > 500 μm
- Maximum load = 500 mN (50.8 g)
4.5 Summary

This chapter listed the types of laser sources and the cutting platform that have been utilised to perform the experimental tasks in the present research work. Different lasers were used with three different range of pulse durations; millisecond, picosecond and femtosecond regimes. The purpose was to understand and study the laser cut qualities with different material removal mechanisms such as melt ejection and ablation. For instance, it is demonstrated that fibre laser cutting in dry condition resulted in poor cut quality. Then wet cutting was undertaken to observe the improvement of quality with the presence of water. In picosecond laser cutting specifically in the UV range, the cutting was performed only in dry condition because the quality obtained here was promising without require the use of water. Femtosecond laser cutting in dry condition resulted in the accumulation of debris within the cutting kerf perimeter as well as recast formation. Hence, water assisted cutting was performed for comparison purposes.
Chapter 5

Comparison of dry and wet fibre laser profile cutting of thin 316L stainless steel tubes for medical device applications

In medical coronary stent fabrication, high precision profile cutting with minimum post-processing is desirable. Existing methods of profiling thin tubular metallic materials are based mainly on the use of Nd:YAG lasers. In recent studies fibre lasers have been used for stent cutting. However, for profiling thin (<4 mm diameter, <200 µm wall thickness) stainless steel tubes, back wall impingements often occur. This chapter presents a comparison of wet and dry pulsed fibre laser profile cutting of 316L stainless steel tubes. When water flow was introduced in the tubes, back wall damage was prevented. Meanwhile, heat affected zone (HAZ), kerf width, surface roughness and dross deposition have also been improved compared with the dry cutting. The scientific study on the effect of internal water flow on laser cutting of thin tubular stainless steel material is reported for the first time.
Chapter 5: Comparison of dry and wet fibre laser profile cutting of thin SS 316L

5.1 Introduction

One of the growing applications of laser micro machining for medical application is the manufacturing of coronary stents. Laser technology is a mostly widely used method in processing stents as compared to electrical discharge machining, water jet cutting, braiding, knitting and photochemical etching [9]. Stent fabrications require a high precision process in order to maintain the slit structures and width. Started with flash-lamp pumped Nd:YAG lasers, laser micro-profiling has been an established tool for coronary stent manufacture. A considerable amount of literature has been published on Nd:YAG laser applications in stent cutting. Kathuria [10] conducted a feasibility study of pulsed Nd:YAG laser precision fabrication of metallic stents. He suggested that a pulsed Nd:YAG laser is a viable tool in creating such fine and mesh structure with slit width of 100 µm. High pulse repetition rate with short pulse duration is preferred for the high cut quality. However, the processed samples were not free from dross and spatter adherence. Work by Raval et al. [119] shows that Nd:YAG laser cutting of stents is associated with slag, oxide layer and unacceptable surface quality.

In the last few years, the emergence of fibre lasers technologies has enabled their increasing applications in medical device micro machining. Kleine and Watkins [108] conducted a comparative study to evaluate the cutting quality between a pulsed lamp pumped Nd:YAG and a fibre laser. Both systems were adjusted to have nearly the same beam quality and beam diameter. The same processing parameters were applied during cutting processes to accomplish a valid comparison results. They have demonstrated that cutting with an Nd:YAG laser slightly degrades the surface quality due to wider striations zone from the top edge of the laser cut. A study by Liu et al. [120] identified that flash lamp pumped Nd:YAG lasers produced large kerf widths and an expansion of heat affected zones due to low facular quality. On the other hand, the poor stability laser outputs of the Nd:YAG laser caused difficulty in reaching small and consistent kerf width in micro machining. Investigation by Meng et al. [96] demonstrated that cutting quality (heat affected zone and average
roughness) was better with a fibre laser compared to an Nd:YAG laser due to single-
mode output and small focused size of the fibre laser. Miller et al. [141] explained
that embrittlement of metal in the heat-affected zone may lead to crack formation and
expansion of the stents which may cause crack propagation and device failure. Thus
significant costs are associated with post processing of Nd:YAG laser machined
stents to produce high quality products. From the aforementioned studies, fibre laser
is seen as a potentially new technology in stent micromachining if heat affected zone,
kerf width and dross could be reduced.

The other issue arises in stent micromachining is the back wall damage either
processed with Nd:YAG or fibre lasers. Preventing back wall damage is a challenge
in cutting such small and thin tubes. The ejected materials from a cut kerf in a form
of hot particles adhere and create backwall damages to the opposite wall of the tube.
Back wall damage would cause rough surface finish and cracks to the stent structures
due to the dissipation of heat by the ejected particles. On the other hand, the laser
beam transmission was also not prevented from reaching the opposite wall causing
deterioration to the backwall.

This chapter aims to investigate the basic characteristics of fibre laser cutting of
stainless steel 316L tube and understand the effect of introducing water flow in the
tubes on minimising back wall damages and thermal effect. The influence of laser
parameters upon cutting quality for fixed gas type and gas pressure was investigated.

5.2 Experimental procedures

5.2.1 The stent cutting system

The tube profiling system used was a Swisstec Micro T15 machine designed for stent
production. This system was integrated with a GSI JK100FL single mode fibre laser
with 100 W peak power. A computer control module was integrated for the process
parameter selection and control. The fibre laser has a 1080 ± 5 nm output wavelength and a beam quality factor, $M^2 < 1.1$. The spot size was measured to be 25 $\mu$m at the focal plane (during the installation by the LPRC experimental officer and machine supplier). The measurement of the spot size was done by firing the laser at low power for a short time to burn a piece of polyimide polymer film. This was done by moving up and down the focusing lens to establish the focal plane. The SEM was used to obtain the spot size dimensions at different focal length.

5.2.2 Materials

In this work, 316L stainless steel with an outer diameter of 3.175 mm and 150 $\mu$m wall thickness was used. The chemical composition for the material is given in Table 5.1.

<table>
<thead>
<tr>
<th>Elements</th>
<th>C</th>
<th>Mn</th>
<th>Cr</th>
<th>Ni</th>
<th>Mo</th>
<th>Si</th>
<th>Cu</th>
<th>N</th>
<th>P</th>
<th>S</th>
<th>Fe</th>
</tr>
</thead>
<tbody>
<tr>
<td>%</td>
<td>0.03 % max</td>
<td>2.0 % max</td>
<td>17.0 %</td>
<td>13.0 %</td>
<td>2.0 %</td>
<td>0.75 % max</td>
<td>0.50 % max</td>
<td>0.10 % max</td>
<td>0.025 %</td>
<td>0.010 % max</td>
<td>Balance</td>
</tr>
</tbody>
</table>

5.2.3 Cutting experiments

The cutting experiments were performed in two cutting conditions, dry and wet by using the pulsed laser mode. Nitrogen was used as an assist gas. Dry cutting has been performed with the presence of the $N_2$ assist gas, and the wet cutting was performed with the presence of an assist gas ($N_2$) and continuous water flow through the inner part of the tube along the tube axis. In the wet cutting, the water pipe with the same diameter of the tube was connected to the tube opening and to the water supply container. In this case, the water flow rate was measured to be 1567 mm$^3$/s which was kept constant during the experiment. Preliminary experiments were carried out to determine the appropriate processing parameters to be used for the comparative
study. The range of parameters chosen was based on necessary average power needed to achieve a full depth penetration for both cutting conditions at the selected cutting speed ranges. Figure 5.1 provides the ranges of average power and cutting speed which resulted in partial and complete cutting for an indication of the process window.

![Figure 5.1: Process window for partial and complete cutting at different ranges of average power and cutting speed.](image)

Table 5.2 shows the ranges of parameters used in the experiment. The parameter variation used was the same for dry and wet cutting condition. The gas pressure was constant at 6 bar as limited by the machine for all the cutting experiment performed. The cutting quality factors investigated were kerf width, surface roughness, dross deposition, back wall damage and heat affected zone (HAZ).
Table 5.2: Cutting parameter ranges.

<table>
<thead>
<tr>
<th>Cutting parameters</th>
<th>Lower Limit</th>
<th>Upper Limit</th>
</tr>
</thead>
<tbody>
<tr>
<td>Peak pulse power (W)</td>
<td>70</td>
<td>100</td>
</tr>
<tr>
<td>Frequency (kHz)</td>
<td>1.5</td>
<td>4</td>
</tr>
<tr>
<td>Pulse width (ms)</td>
<td>0.1</td>
<td>0.2</td>
</tr>
<tr>
<td>Cutting speed (mm/min)</td>
<td>250</td>
<td>2000</td>
</tr>
</tbody>
</table>

In this study, the computer aided design (CAD) data of the cut profiles was created and transferred to the Micro T15 computer system. The data were translated into G-code programming by the computer system. In order to study the quality characteristics of the fibre laser stent cutting system, a simple geometry was designed to cut the stainless steel 316L tubes. Initially, the tubes were cut into two separated parts; Part A and part B as shown in Figure 5.2. The simple geometry was used to assess the basic characteristics including kerf width, surface roughness, dross deposition, back wall damages and HAZ. A more complex profile was used to assess the heat effects, particularly along the cut kerf.

![Part A and Part B](image)

Figure 5.2: Scanning electron microscope (SEM) image of the laser cut geometry.
5.3 Results

5.3.1 Effects of cutting parameters upon kerf width and surface roughness

Figure 5.3 shows the relationship between the kerf width and laser cutting parameters. The standard deviations were taken to produce the data error bars. The results show that the kerf width increased as the peak pulse power, frequency and pulse width increased (Figures 5.3a, 5.3b, and 5.3c). During the cutting process, the average power supplied was controlled by these three parameters. The small variation in average power resulted in a large variation of kerf width. Increasing the speed led to reduction of kerf width after a critical cutting speed was achieved (Figure 5.3d). Less interaction time between the laser beam and material reduced the energy supplied to the cut kerf, and thus less amount of material was melted. The kerf width size variation was investigated by varying the speed between 250 mm/min up to 2000 mm/min (maximum equipment speed limit). Critical cutting speed for dry cutting and wet cutting were 1250 mm/min and 1000 mm/min respectively where the kerf width started to decrease after these particular points. The kerf widths obtained in this experiment were within the range of 28 ~ 40 μm. Minimum kerf was obtained at high cutting speed, 1750 mm/min and 2000 mm/min. At this point, the highest table speed programmed may not have been realised due to the acceleration-distance limitation of the cutting stage. Hence, the results in Figure 5.3d are not comparable for 1750 and 2000 mm/min. This suggests that, for the machine capability and component size, the speed of 1750 mm/min appears to be the maximum that can be realised. From Figure 5.3, it can be clearly seen that the wet cutting produced a low kerf width compared to dry cut. In addition, it is seen that kerf width increases with the repetition rate in the fibre laser cutting as more energy was delivered to the process.
Figure 5.3: Kerf width as a function of laser cutting parameters in dry and wet cutting (a) peak pulse power (b) frequency (c) pulse width (d) cutting speed.

Figure 5.4 shows the influences of laser parameters upon surface roughness. The surface roughness increased with the increasing peak pulse power. However, the surface roughness reduced when the peak pulse power increased from 90 W to 100 W (Figure 5.4a). Higher pulse frequency improves surface quality as shown in Figure 5.4b. Pulse width significantly affects the surface roughness with a rougher cut surface produced as the pulse width increases (Figure 5.4c). From Figure 5.4d, surface roughness decreased as the cutting speed increased until 1250 mm/min for the dry cut and 1000 mm/min for wet cut. After these points, the surface roughness gradually increased and were more pronounced at higher cutting speed for wet cutting. This result shows that more power is required after certain point of speed to improve surface quality. In comparison with dry cutting, wet cutting led to better surface quality.
5.3.2 Dross formation

The dross formation after the laser cutting in dry and wet cutting condition was observed as shown in Figure 5.5. From the observation, the processed samples were not free from dross. In both cutting conditions, the dross deposition was heavy at the low cutting speed of 500 mm/min and slightly reduced at 1000 mm/min. The molten material was not totally ejected out from the cut kerf and was attached to the bottom side of the cut wall. Above 1000 mm/min, the dross formation is not significant. By comparing the dross deposition at cutting speed 1000 mm/min for both wet and dry cutting conditions, dross was significantly reduced in the wet cutting condition at 1000 mm/min. In inert gas cutting, the energy solely came from the focused laser beam. The presence of inert gas does not contribute any additional energy to the cutting point (cooling of the cut zone may be resulted) thus producing high viscosity and high surface tension of molten material. The obtained results showed that 6 bar gas pressure was not enough to act as a mechanical force to drag away the molten...
material. The presence of water was not sufficient to clean the dross. Water flow, however, helped to achieve a reduction in dross. It is shown that higher pressure inert gas is required to clean the dross in achieving high cutting quality.

<table>
<thead>
<tr>
<th></th>
<th>500 mm/min</th>
<th>1000 mm/min</th>
</tr>
</thead>
<tbody>
<tr>
<td>Dry Cutting</td>
<td><img src="image1.png" alt="Image" /></td>
<td><img src="image2.png" alt="Image" /></td>
</tr>
<tr>
<td>Wet Cutting</td>
<td><img src="image3.png" alt="Image" /></td>
<td><img src="image4.png" alt="Image" /></td>
</tr>
</tbody>
</table>

Figure 5.5: Dross deposition at different cutting speed in different cutting condition.

### 5.3.3 Back wall damage

Preventing back wall damage is a challenge in cutting small and thin tubes. The ejected materials from a cut kerf in a form of hot particles adhered and created backwall damages to the opposite wall of the tube. Back wall damage would cause rough surface finish and cracks to the stent structures due to the dissipation of heat by the ejected particles. On the other hand, the laser beam transmission was also not prevented and transmitted to the opposite wall causing deterioration to the backwall. In the dry cutting, it can be observed that the undesired particles were scattered on the opposite wall and this phenomenon was not avoidable by changing cutting parameters (Figure 5.6a). Heat contained in these particles was transferred to the back wall. In the wet cutting, a clean and spatter free back wall was obtained as can be seen in Figure 5.6b. The continuous water stream carries away the hot particles after they are ejected from the cut kerf and minimizes the heat transfer to the back wall. Figures 5.7a and 5.7b compare the surface profile analysis of the back wall for both wet and dry cutting conditions. The wet cutting seems to be effective in preventing the backwall damage.
5.3.4 Heat effects

Heat effect on the surrounding material is a critical factor in cutting thin materials especially in medical device application. Small and thin materials are very sensitive to thermal distortion. Experiment results show that shorter pulse width and low average power reduced the thermal distortion. A shorter pulse width is always
recommended in cutting materials sensitive to distortion. At a high average power (40 W), dry cutting (Figure 5.8a) resulted in a noticeable thermal effect and surface oxidation along the cut while wet cutting (Figure 5.8b) reduced this effect even at high average powers. A more complex profile similar to a medical stent was used to assess the heat effect particularly along the cut kerf. Figure 5.9 shows the thermal discoloration in dry stent cutting, while cutting in the water assistance condition gave a clean and bright surface. Wet cutting relatively created controllable heating during the cutting process and reduced the extent of heat significantly that benefits particularly in cutting thin and small diameter tubular materials.

Figure 5.8: Comparison of thermal effect along the cut kerf at 40 W average power (a) dry cutting (b) wet cutting.

Figure 5.9: Thermal discoloration in stent cutting before removing the excess material (a) dry cutting (b) wet cutting.

The heat affected zones (HAZ) corresponding to a high average power (40 W) for both cutting conditions were compared after the laser cut samples were undergone a
series of grinding, polishing and etching processes. The grain structures were revealed by immersing the samples in Glyceregia solution for 60 seconds. The average laser power of 40 W was selected as a comparative parameter to observe the effectiveness of water to reduce the heat affected zone even at high power input. As expected, wet cutting resulted in less pronounced HAZ with the width less than 10 µm while dry cutting reached 30 µm (Figure 5.10).

Figure 5.10: Tranverse section of the cut kerf showing the changes of the microstructure associated with HAZ when cutting at high average power (40W) (a) dry cutting (b) wet cutting.

5.4 Discussions

In fibre laser cutting processes, many factors affect the cutting quality. To understand the effect of each processing parameter, the approach of varying one parameter at a time and keeping the other parameters constant was used. In this experiment, the varied parameters were peak pulse power, pulse width, frequency and cutting speed. The average power, $P_{ave}$ can be obtained from:

$$P_{ave} = P_p \tau f$$

(5.1)

Where, $P_p$ is the peak pulse power, $\tau$ is pulse width and $f$ is pulse frequency.

Increasing all these three parameters ($P_p$, $\tau$, $f$) will lead to the increase of laser average power delivered to the cutting process. Higher average power generates
more energy transfer to the kerf and increases the amount of melted material which translates to a wider kerf. Increasing the pulse width causes a larger kerf due to more material being melted and removed by the gas. As the cutting speed reduces, the interaction time between laser beam and material increases which creates a larger kerf. Increasing the peak pulse power and pulse width resulted in increasing surface roughness. Higher frequency applied to the cutting process improves surface quality due to high pulse overlapping. Kerf width and surface roughness results seem to indicate that pulse width is a dominant factor affecting the cutting quality. The findings from this study suggest that the introduction of water flow inside the tube reduced the local heating during the cutting process thus minimized the amount of melted material and produced smaller kerfs. Additionally, the presence of water has affected the melt flow behaviour resulting in reduced surface roughness compared to that during the dry cut.

In this experiment, with the flowing water, a clean back wall and low thermal effect even at high average powers and high pulse energy was demonstrated. Figure 5.11 illustrates the mechanism of molten material behaviour after being ejected from cut kerf in dry and wet cutting. The molten metal solidifies and forms undesired spatter on the back wall in the case of dry cutting (Figure 5.11a). This created damages to the opposite wall as the wall thickness is thin and thus thermal conduction is not as efficient as in case of thick walls. Figure 5.11b shows the clean and spatter-free back wall that was obtained by using the wet cutting. Water flow from the inner diameter of the tube ideally removed all the molten materials before they permanently adhere to the back wall.
Significant improvements with wet cutting to minimize the heat effect and back wall damage were achieved in this work. As every single stent strut needs to be cut without any defects, the water preventive method shows promising results in reducing the heat effect and backwall damages. This is due to the water stream that carries away the hot debris which minimizes the heat transfer and impingement to the backwall.

In order to investigate effect of the water in attenuating the laser transmission during the wet cutting, water transmission spectrum was measured for the 900-1100 nm wavelength at every 1 nm by using an Ocean Optics spectrometer (Figure 5.12). The measurement was made with a water path length of 1 cm. The relationship between light transmission and absorbance can be obtained based on Beer-Lambert Law as referred in [142].

\[ A = -\log_{10} T \]  \hspace{1cm} (5.2)

where, \( A \) is absorbance and \( T \) is transmittance. The absorption coefficient, \( \alpha \) is defined by:
\[ \alpha = \left( \frac{1}{x} \right) \ln \left( \frac{1}{T} \right) \]  
\hspace{1cm} (5.3)

Here, \( x \) is the water path length. The absorption length, \( z \) therefore is related to the absorption coefficient, \( \alpha \) and is given by

\[ z = \frac{1}{\alpha} \]  
\hspace{1cm} (5.4)

From equations (5.2), (5.3) and (5.4), the water absorption spectrum was calculated and plotted based on the transmission data obtained from spectrometer. Through the above calculations, the light absorption length for water media is around 4.2 cm for fibre laser wavelength (\( \lambda = 1080 \pm 5 \) nm) as shown in Figure 5.13. The absorption spectrum for this spectrum band is in agreement to the results shown by Kruusing [143].

Figure 5.12: Water transmission spectrums recorded by spectrometer for 1 cm water path length.
The water thickness flowing inside the tube is $x = 0.2875$ cm. Based on the Beer-Lambert law, the fraction of the light absorbed by each layer of solution is proportional to the path length. Thus, the transmission for water path length, $x = 0.2875$ cm has been computed based on the spectrometry analysis. The transmission obtained for this particular water path length is 94%, which indicates that most of the beam intensity strikes the backwall and showing that water does not absorb much laser intensity. This elucidates that the transmissivity and absorption of light in water is associated with the light path length in water. High absorptivity of laser energy could not be expected in a thin water path length. However, the transmission is expected to be less than 94% due to existence of molten material (produced during cutting) in the water that slightly increases the attenuation.

The spectrum analysis shows that water does not play a significant role in attenuating the laser beam in cutting small diameter tube since the 100% absorption can be obtained at around 4.2 cm of water depth. In wet cutting, water played an important role in cooling the workpiece and carries away the debris rather than attenuating the beam. A constant rate of water circulation removes local heating of water adjacent to the cutting surrounding and improves the cooling. It is well known that water cools the cutting zones more rapidly than particular gases or gas mixtures such as air [143].
In the current work, the wet cutting results obtained were not free from dross attached to the cutting edge. A conceivable reason was the inadequate inert gas pressure which is not sufficient to force away the molten material from the cutting area in preventing dross.

5.5 Conclusions

The experimental results comparing dry and wet stainless steel 316L tube cutting were reported. Wet cutting enabled significant improvement in cutting quality. Wet cutting resulted in narrower kerf width, lower surface roughness, less dross, absence of back wall damages and smaller HAZ which would lead to reducing the cost of post processing. Laser average power and pulse width play a significant role in controlling the cutting quality. Increasing the pulse width increased beam/material interaction time which increased the kerf width and surface roughness. Wet cutting is ideally suitable for cutting thin wall and small diameter tubes specifically in stent production.
Chapter 6

Picosecond laser micromachining of nitinol and platinum iridium alloy for coronary stent applications

The demand for micromachining of coronary stents by means of industrial lasers rises quickly for treating coronary artery diseases which cause more than one million deaths each year. The most widely used type of lasers for stent manufacturing are Nd:YAG laser systems with a wavelength of 1064 nm with pulse lengths of $10^{-3}$-$10^{-2}$ seconds. Considerable post processing is required to remove heat affected zones (HAZ), and to improve surface finishes and geometry. Using a third harmonic laser radiation of picosecond laser ($6 \times 10^{12}$ s pulse duration) in UV range, the capability of the picosecond laser micromachining of nitinol and platinum-iridium alloy for coronary stent applications are presented. In this study dross-free cut of nitinol and platinum-iridium alloy tubes are demonstrated and topography analysis of the cut surface is carried out. The HAZ characteristics have been investigated by means of microscopic examinations and measurement of micro-hardness distribution near the cut zones.
6.1 Introduction

Stents are typically made from biocompatible materials such as stainless steel, nitinol (Ni-Ti alloy), cobalt-chromium, titanium, tantalum alloys, platinum-iridium alloy as well as polymer [95]. The most widely used material for self expandable stents is nitinol [9]. Self-expanding stents are advantageous in that less damage to the vessel is caused, due to low pressure while it is dilated. Moreover, it is beneficial for complex lesions as they can be securely implanted without the use of a balloon for expanding purpose [144].

As the complex design of stent requires precise manufacturing processes, a large volume of work have been undertaken in order to supply reliable and trustworthy stent devices for implantation. Laser stent manufacturing started more than ten years ago and has become the primary choice in stent cutting applications. Starting with the application of lamp-pumped Nd: YAG lasers (wavelength of 1,064 nm, and a pulse duration in the millisecond range), the laser technology rose quickly to suit the demands in cutting such intricate shapes with a development of ultra-short pulse laser (picosecond and femtosecond pulse duration). Ultra-short pulse has been an interest due to its potential to minimise and eliminate the heat affected zone. Recently, picosecond laser have been demonstrated to micromachine metals and other industrial materials with the same precision and similar removal rate per pulse. Picosecond pulse are short enough to avoid thermal diffusion of energy and reach the peak power densities necessary for high precision machining processes [129].

The ablation and machining of metal sheets using picosecond lasers have been investigated by several groups [123, 124, 130]. However no specific study has been reported regarding picosecond laser cutting of nitinol and platinum-iridium alloy tubes particularly for stent applications. The novelty of this research is to present the remarkable achievement of picosecond laser in manufacturing stents. This contribution, concentrated on the capability of picosecond laser micromachining of nitinol and platinum-iridium alloy in improving the cut quality.
6.2 Experimental Procedures

6.2.1 Materials

The experimental materials were nitinol tubes, (equatomic composition of nickel and titanium) and platinum-iridium alloy (Pt-10% Ir), commonly used for the manufacture of medical devices and surgical implants. The nitinol tube has an outer diameter of 5.0 mm and 280 µm wall thickness, while the platinum-iridium alloy tube has a 1.88 mm outer diameter with 67 µm wall thickness. Nitinol is a well-known shape memory alloy and an established stent material that has been used as a self-expanding stent. Platinum-iridium alloy is a precious metal with a high degree of biocompatibility and biofunctionality which has recently been used as a stent material.

6.2.2 Equipment

Picosecond laser radiation was carried out by a Trumpf TruMicro 5350 laser with \( \lambda = 343 \) nm (third harmonic generation of wavelength), a beam quality factor \( M^2 = 1.2 \) with a Gaussian beam intensity profile. Theoretical spot size at focus was determined to be 5.2 µm. A high precision Swisstec Micro T-15 delivery system was utilised in this experiment. This machining unit was designed purposely for stent production.

Characterization of the laser cut samples was conducted by scanning electron microscopy (Hitachi S-3400N) for high resolution imaging. Cut edges were characterised by a MicroXam white light interferometer surface profile measurement system. To analyse the striations, 3D cut edges have been reconstructed by using MeX software package (Alicona Imaging GmbH), from three different tilted SEM images which allow the measurement of striations along the defined paths of the 3D image. The hardness value was measured using a MTS Nano-indenter XP (Agilent). Seven indentations were made for each specimen with three repetitions. The base material hardness was measured as a reference.
6.2.3 Laser Cutting

Prior to laser cutting, the desired cut geometry was created with computer aided design (CAD) software. The CAD data was then converted into ISO file which enables the machining unit computer module to translate it into a G-code program. The tube sample was loaded into the workstation horizontally and was guided by a bushing mechanism. The focal position of the laser was set at 20 µm below the top surface. During the cutting process, the experimental tube was rotated and moved in the transverse axes directed by the G-code movement while the laser remains stationary. The schematic diagram of laser tube cutting process is represented in Figure 6.1. The process parameters were average power, repetition rate, pulse duration, cutting speed and assist gas pressure. Preliminary experiments were carried out to determine the appropriate processing parameters to be used for further investigation. The chosen average power, cutting speed and repetition rate were based on the sufficient energy needed to cut both nitinol and platinum-iridium alloy with minimum passes. By varying pulse durations (other process parameters were kept the same), 6 ps pulse duration was found to be ideal for use which resulted in clean and fine cut quality. The pulse duration selected is in agreement with the previous work by Dausinger et al. [129], who suggested that a pulse duration between 5 to 10 ps was optimal for micromachining of metals. The set of process parameters for both materials are shown in Tables 6.1 and 6.2. The advantages of the 6 ps UV pulses (λ=343 nm, used in this work) is the high photon energy that provides better absorption compared to the infrared photons. Laser beam of UV wavelength also enable the beam to be focused to a smaller spot size (the spot size in this work is 5.2 µm), compared to the work by Li et al. [100] using 100 fs, 800 nm pulses where the spot size was >20 µm. The 6 ps UV pulses give much better focusing suitable for micromachining delicate and thin materials in coronary stent applications.
Chapter 6: Picosecond laser micromachining of nitinol and platinum iridium alloy

Figure 6.1: Schematic diagram of laser tube cutting.

Table 6.1: Process parameters for machining nitinol.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Laser wavelength</td>
<td>343 nm</td>
</tr>
<tr>
<td>Average power</td>
<td>10 W</td>
</tr>
<tr>
<td>Repetition rate</td>
<td>200 kHz</td>
</tr>
<tr>
<td>Pulse duration</td>
<td>6 picosecond</td>
</tr>
<tr>
<td>Cutting speed</td>
<td>25 mm/min</td>
</tr>
<tr>
<td>Assist gas</td>
<td>4.5 bar Argon</td>
</tr>
<tr>
<td>Number of passes</td>
<td>3</td>
</tr>
</tbody>
</table>

Table 6.2: Process parameters for machining platinum-iridium alloy

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Laser wavelength</td>
<td>343 nm</td>
</tr>
<tr>
<td>Average power</td>
<td>7 W</td>
</tr>
<tr>
<td>Repetition rate</td>
<td>200 kHz</td>
</tr>
<tr>
<td>Pulse duration</td>
<td>6 picosecond</td>
</tr>
<tr>
<td>Cutting speed</td>
<td>100 mm/min</td>
</tr>
<tr>
<td>Assist gas</td>
<td>4.5 bar Argon</td>
</tr>
<tr>
<td>Number of passes</td>
<td>3</td>
</tr>
</tbody>
</table>
6.3 Results

6.3.1 SEM images of the laser cut samples

The tube cutting with the picosecond laser by using aforementioned laser parameters were demonstrated for the nitinol and the platinum-iridium alloy. The samples were examined using SEM directly after cutting process without undergoing any post treatments. The high cut quality geometry was produced as shown in Figure 6.2.

![Figure 6.2: SEM images of laser cut samples: (a) nitinol and (b) platinum-iridium alloy.](image)

6.3.2 Demonstration of dross free cut

As found through the experimental work, the cutting process with 6 ps pulse duration (for both samples) succeeds in achieving dross free cut with sharp edges. No post-processing was undertaken to the processed samples after cutting process. This clearly demonstrates that dross-free cut is feasible with the picosecond laser cutting. In this process, the ablation of material is considered to be dominated by rapid vaporisation. The energy is deposited rapidly to the material by ultra-short pulses. The material is heated to the vaporisation state rapidly which reduced the molten volume in the laser/material interaction zone. In addition, the use of argon as an assist gas significantly reduced the redeposition of the ablated material to the parent material. This can be explained due to argon being an inert gas with extremely low level of reactivity and does not supply any additional heat to the process while high density of argon giving high mechanical force effectively blowing away the ablated
material. During this process, it should be noted that a defocused beam with the focal plane 20 µm under the top surface was used in order to obtain high cut quality. Considering these, high quality cutting with no dross was achieved during the process. This result shows the feasibility of picosecond laser cutting to meet the high quality surface demand in stent manufacture. High resolution SEM images at different magnifications showing the dross free cut for both samples are displayed in Figures 6.3 and 6.4.

![Figure 6.3: Nitinol cut edge without any post processing showing the dross free cut and sharp edge in different magnifications: (a) 60x and (b) 200x.](image)

![Figure 6.4: Platinum-iridium alloy cut edge without any post processing showing the dross free cut and sharp edge in different magnifications (a) 60x (b) 400x.](image)
6.3.3 Topography analysis of the cut surface

6.3.3.1 Surface roughness

Analysis was carried out to characterise the topography of the cut edges to determine the surface finish quality. The average roughness of the nitinol cut surface was found to be $R_a=1.34 \, \mu m$, as shown in Figure 6.5. In comparison, platinum-iridium alloy produced a better surface quality with $R_a=0.49 \, \mu m$ as shown in Figure 6.6.

Figure 6.5: Nitinol cut surface tomography: (a) Cut edge, (b) Enlarged part of the analyzed cut edge and (c) Interferometer image giving the surface profiles data.

Figure 6.6: Platinum-iridium alloy cut surface tomography: (a) Cut edge, (b) Enlarged part of the analyzed cut edge and (c) Interferometer image giving the surface profiles data.
6.3.4 Striations formation on the cut surface

Striation refers to periodic lines appearing on the cut surface. Striation mainly affects the surface quality of the cut zones. To analyse the striations, 3D cut edges were reconstructed by using the MeX software package in the SEM system as shown in Figures 6.7a and 6.8a. These periodic lines reflect the effect of laser beam moving along the surface, generating pressure gradients across the cut kerf and varying vaporisation fronts. This phenomenon created regularly spaced striations and increased the surface roughness on the cut surface. From the result, it is clearly seen that striations occurred in both samples. The striation profiles were measured along the upper (L1) and the lower (L2) of the cut surface within 90 µm path length for both samples as represented in Figures 6.7 and 6.8. Nitinol cut surface demonstrated deep and pronounced striations with the maximum depth around 3 µm. It is found that the striation depth is higher in the upper part of the cut kerf and slightly lower towards the bottom part which is dependent on the thickness of the molten layer. The striations frequency is higher at the bottom part as compared to the upper part.

![Figure 6.7: 3D image of the nitinol cut surface reconstructed by MeX software package. L1 and L2 represent the path length for striations measurement, (b) Striations profile across the cut edge upper part, L1, (c) Striations profile across the cut edge lower part, L2.](image)

In contrast to nitinol, platinum-iridium alloy exhibited a very shallow and uniform striations pattern. The striations depth is found to be less than 0.4 µm. High striation...
frequency has been observed on the platinum-iridium alloy surface that created higher surface quality as can be observed in Figure 6.8. Theoretically, if the frequency is increased indefinitely, the geometric surface roughness should go to zero [47]. Additionally, the striations frequency increased with cutting speed and this was observed in machining both materials. It is also reported in literature that light emitted from the cut front correlates with the striation frequency [145].

![Figure 6.8 : 3D image of the platinum-iridium alloy cut surface reconstructed by MeX software package. L1 and L2 represent the path length for striations measurement, (b) Striations profile across the cut edge upper part, L1, (c) Striations profile across the cut edge lower part, L2.](image)

### 6.3.5 HAZ analysis

In laser cutting, heat generated by the energy source can diffuse and alter the metallurgical properties and varies the hardness of the neighbouring material near the heated zones. Changes in hardness in the cut zone indicate the recrystallization that takes place within this area, known as HAZ. Typical procedure in characterising HAZ involves revealing the grain structure by means of chemical etching. However, according to work by Huang et al. [46], etching hot worked nitinol does not reveal clear grain boundary for accurate measurement of HAZ. In addition, the solution used to reveal the grain structure of nitinol involves hydrofluoric acid (HF) which is highly hazardous. On the other hand, hardness measurement is also important to be
considered as changes in material hardness as the result of microstructure changes and it could cause stents to rupture during implanting in human body. This makes nano-indentation hardness an ideal method to measure the HAZ for both samples.

Hardness for nitinol and platinum-iridium alloy samples were obtained as shown in Figures 6.9 and 6.10 respectively. The hardness values for both samples were recorded 35 µm along the path from and perpendicular to the cut edges. The measurement was taken around 5 µm from the cut edge with 5 µm displacement afterwards. It was observed that for nitinol the hardness within 10 µm distance of the cut surface was on average 6.35 GPa which is harder compared to the average hardness of the base material (3.7 GPa). This layer was hardened due to recast layer formed on the cut surface. This shows that the recast layer transferred the excessive heat to the bulk material underneath around 5 µm depth. Due to the activation of thermal energy, the martensitic transformations occurred that resulted in the changes of hardness. At the distance of 15 µm and downwards the average hardness profile was found to be uniform (around 3.6 ~ 3.8 GPa) which is comparable to the parent material hardness. There is no material metallurgical change beyond 15 µm from the cut edge.

Platinum-iridium alloy hardness appeared to be consistent along the 35 µm path from the cut edges. The measurement was taken around 3 µm from the cut edge with 5 µm displacement afterwards. Average hardness of platinum-iridium alloy was found to be around 3.2 ~ 3.4 GPa which is similar to the parent material hardness (3.2 GPa). The result shows that there is no detectable modification in the hardness of the material. The phenomena can be explained by the fact that the short pulsed laser removes material rapidly without diffusing excessive heat to the base material. Consequently no significant recrystallization takes place. The major difference between nitinol and the platinum-iridium alloy is their thermal conductivity. Platinum-iridium alloy (Pt- 10%Ir) has a thermal conductivity of 60~72 Wm⁻¹K⁻¹ [146] while nitinol has the much lower thermal conductivity of 10~18 Wm⁻¹K⁻¹ [147]. Higher thermal conductivity of the platinum-iridium alloy allows the heat to be dissipated more quickly. In addition, the phase transformation temperature of
nitinol is much lower than that of platinum alloy that makes platinum less vulnerable to phase transformations.

Figure 6.9: Hardness test at various distances from cut edge for the nitinol sample: (a) measured surface and (b) hardness value.

Figure 6.10: Hardness test at various distances from the cut edge for the platinum-iridium alloy sample, (a) measured surface and (b) hardness value.
6.4 Discussions

In case of ultra-short laser pulse (pico and femto time range), the material removal process is dominated by material vaporisation which is different from longer pulse and continuous wave laser cutting which would involve considerable melting and melt removal by a gas jet. The ultra-short pulse laser cutting process involves rapid energy deposition to the material which causes material to have rapid temperature rise to the vaporisation state. Within the short interaction time at 6 ps pulse duration and pulse energy up to 50 micro joules contained in a single pulsed emission, only the electrons within the material was heated during the assigned pulse duration. Once the laser pulse has stopped, the lattice of the material experiences the influence of the overheated electrons which allow high precision and minimal heat influence within the materials as explained by Meijer et al. [74]. In this process, the amount of molten material is negligible. This can be characterised through the clean backwall from the debris/spatter adherence as can be seen in Figure 6.11. There is no trace of the melted material attached to the back wall. The remarkable cutting achievement (dross free and clean backwall) is very important in micro cutting for medical implants since the requirements for this application is very strict in quality. The cut without residues and a clean backwall free from spatter implies that minimal heat diffusion with rapid solidification occur during machining with ultra-short pulse laser.

Figure 6.11: SEM micrograph showing inner wall of nitinol is clean from the debris/spatter adherence.
In this investigation, low material thermal conductivity influenced the HAZ formation for nitinol. Thermal conductivity for nitinol is regarded as poor with a value of $10\sim18 \text{ Wm}^{-1}\text{K}^{-1}$ [147]. When the material is irradiated, the temperature rise is instantaneously confined to the region being heated. The heat generated tends to localise the heat and extend the HAZ due to the small heat conduction rate. Also nitinol has very low phase transformation temperature (around $60^\circ\text{C}$) [148] which easily affect the microstructure formation. A different phenomenon occurs in high thermal conductivity platinum-iridium alloy material where the heat generated is rapidly conducted away from the cut region preventing excessive heating to the neighbouring material resulted no HAZ. Platinum-iridium alloy has much higher thermal conductivity with value $60\sim72 \text{ Wm}^{-1}\text{K}^{-1}$ [146]. High thermal conductivity can act as a heat sink, drawing heat away from a cut zone and cooling at an accelerated rate. Kannatey-Asibu [44] stated that a higher thermal conductivity for a material, makes the isotherms more circular and reducing the temperature gradient in front of the heat source. Thus, the heat is more concentrated on the cut zone and created a local heating reducing the extent of HAZ. In addition, the phase transformation temperature of platinum-iridium alloy is around $1000^\circ\text{C}$ [149] which is over 15 times higher than that of nitinol. Therefore less heat affected zone was observed for platinum-iridium alloy cutting.

In further studies by authors, it has been found that the picosecond laser offers a great potential in machining other stent materials including cobalt chromium, silver and titanium as shown in Figure 6.12. Similar to the earlier results, the dross free cut has been achieved for the other stent materials by using the 6 ps pulse width. The benefit with the picosecond laser micro-machining is that the material distortion can be limited which allow one to cut more intricate shapes and tight tolerances could be achieved as compared to a long pulses laser. In addition, the material exposed to picosecond pulses can directly burst into vapour and plasma which has resulted in the dross free cut. The picosecond laser also causes a minimal substrate heating, demonstrating a great potential for fabricating materials with low melting point such as polymer with the excellent cutting tolerances as shown in Figure 6.13. For
polymer machining, photo-chemical absorption may take place as a result of high photo energy of the UV beam and the low molecular bond energy of the material resulting in a “cold machining” without going through a thermal heating process. The short pulse laser ablation phenomenon benefits in structuring polymer as the material properties are maintained.

Figure 6.12: SEM images showing a great potential of pico second laser in cutting other stent materials, (a) and (b) cobalt chromium, (c) silver, and (d) titanium.
Laser cutting of stents also have been done widely by the other investigators using longer pulse duration lasers. The cutting results showed the formation of dross and spatter adherence in the inner walls [10, 101, 118, 119, 150]. They applied acid pickling as an initial post processing after laser cut to remove dross, debris and spatter adherence by means of chemical attack. In this process, the laser cut stent was immersed in acidic solutions from 15 to 60 minutes at an optimum temperature. These defects can be removed by means of acid pickling, but the nature of the process involving material removal resulted weight loss and strut width reduction which affected stent accuracy. Zhao et al. [151] reported that acid pickling reduced 7.5 % of stent weight and 5.41 % of width reduction. The pickling process should be controllable to minimise excessive removal of stent material. The use of picosecond laser cutting is an exception for the pickling process. The cut without residue minimises the need for a secondary process which is hazardous and time consuming. Table 6.3 compares cutting quality obtained with picosecond laser and longer pulses laser (Nd:YAG and fibre lasers) from literature in processing biocompatible materials, where the advantages of ultra-short pulse picosecond laser can be seen.
Table 6.3: Comparison of cutting quality between short and long pulses laser [118, 119, 150].

<table>
<thead>
<tr>
<th>Laser type</th>
<th>Wavelength</th>
<th>Material</th>
<th>Machining Condition</th>
<th>Cut quality</th>
</tr>
</thead>
</table>
| Nd:YAG laser     | 1064 nm    | SS 316L tube 1.72 mm outer diameter 0.11 mm wall thickness | Average power: 2.5 W  
Frequency: 50 Hz  
Cutting speed: not provided  
Pulse width: 0.15 ns | •Formation of oxide layer and slag  
•Unacceptable surface quality  
•HAZ ~ detectable [119] |
| Fibre laser      | 1080 nm    | SS 316L tube 3.175 mm outer diameter 0.15 mm wall thickness | Average power: 15 ~ 40 W  
Frequency: 1.5 ~ 4.0 kHz  
Cutting speed: 250 ~ 2000 mm/min  
Pulse width: 0.13 ns | •Burs, dross and backwall damages  
•Ra = 0.34 µm  
•HAZ = 30 µm [150] |
| Picosecond laser | 343 nm     | Nitinol tube 5.0 mm outer diameter 0.28 mm wall thickness | Average power: 10 W  
Frequency: 200 kHz  
Cutting speed: 25 mm/min  
Pulse width: 6 ps | •No dross, smooth edges and clean backwall  
•Ra = 1.34 µm  
•HAZ = 15 µm This study |
|                  | 343 nm     | Platinum alloy tube 1.88 mm outer diameter 0.067 mm wall thickness | Average power: 7 W  
Frequency: 200 kHz  
Cutting speed: 100 mm/min  
Pulse width: 6 ps | •No dross, smooth edges and clean backwall  
•Ra = 0.49 µm  
•No present of HAZ |

In this work, the cutting quality achieved could reduce the post processing requirement as the dross and spatter are negligible. The surface quality for nitinol requires improvement since the $R_a = 1.34 \, \mu m$ while platinum-iridium alloy exhibits very fine surface quality, $R_a = 0.49 \, \mu m$. Considering the cutting process uses an inert gas, it can be said that the main contributor to the surface roughness value is the striations due to hydrodynamic instabilities of the molten front. As proposed by Shuocker [152], this phenomenon happens as a result from small fluctuations in laser power and gas flow which contribute to their formation.

The findings from this study suggest that the picosecond laser operated in UV wavelength regime with an advantage of high photon energy gives a very promising cutting quality, particularly for the platinum-iridium alloy and polymer. Over $10^8$ reductions in pulse length as compared to µs and ms pulsed lasers created a new laser micro-machining opportunity that allows the manufacturers to explore the benefits of
new phenomenon in laser micro machining that combines fine ablation capability with an acceptable material removal rate. Picosecond pulse laser machining demonstrated the capability of producing stents with a similar precision to those produced by femtosecond laser [128]. However, at this stage without having optimised the processing parameters, heat affected zone is detectable on nitinol cut edge and the surface quality requires improvement. By optimising the processing parameters (importantly repetition rate and cutting speed), a fine surface quality with no HAZ is expected to be achieved which would require no post-processing. In most cases, repetition rate and cutting speed are important parameters affecting the cut quality in terms of surface quality and HAZ. Generally, high repetition rate produces better surface quality due to high pulse overlapping. High cutting speed has the potential to minimise thermal input to the workpiece due to less material and beam interaction [150].

6.5 Conclusions

A remarkable achievement in micromachining nitinol and platinum-iridium alloy tubes at 6 ps pulse duration has been reported. The mechanism of ultra-short pulse laser allows phenomenon of rapid phase transition from solid to gas with insignificant amount of liquid phase. As beam and material interaction is within the pico second time range, the heating directly bursts the solid material into vapour and plasma that promote dross free cutting. The main achievement in this study is dross-free cut as well as clean back wall which may eliminate the need for extensive processing after the laser cut to meet industrial yield. The observed HAZ (around 15 µm) for nitinol is much less than that processed with other long pulses lasers. Negligible HAZ for platinum-iridium alloy has been achieved. Nitinol exhibits a rougher surface due to the recast layer and formation of striations with an $R_a=1.34$ µm, while platinum-iridium alloy exhibits very fine surface quality with $R_a=0.49$ µm. Large differences in the material properties (transformation temperature and thermal conductivity) between the two materials would be the main causes for the observed
difference in heat affected zones. Results obtained from this study show that picosecond laser is a viable tool inviting for machining delicate and sensitive materials for coronary stent applications.
Chapter 7

Underwater femtosecond laser micromachining of thin nitinol tubes for medical coronary stent manufacture

Micro-profiling of medical coronary stents has been dominated by the use of Nd:YAG lasers with pulse lengths in the range of a few milliseconds, and material removal is based on the melt ejection with a high pressure gas. As a result, recast and heat affected zones are produced, and various post-processing procedures are required to remove these defects. This chapter reports a new approach of machining stents in submerged conditions using 100 fs pulsed laser. A comparison is given on dry and underwater femtosecond laser micromachining technique of nickel-titanium alloy (nitinol) typically used as the material for coronary stents. The characteristics of laser interactions with the material have been studied. A femtosecond Ti:Sapphire laser system (wavelength of 800 nm, pulse duration of 100 fs, repetition rate of 1 kHz) was used to perform the cutting process.
Chapter 7: Underwater femtosecond laser micromachining of thin nitinol tubes

7.1 Introduction

Laser technology has been widely applied for processing nitinol, particularly for medical applications. Nd:YAG lasers have dominated stent manufacture where the mechanism of material removal is mostly thermal, based on melt ejection [10, 119, 120]. As a result, recasts and heat affected zones (HAZ) are prominent, and post processing procedures are required to remove these defects. Over the last decade, ultra-short pulse (picosecond to femtosecond) lasers have been available for industrial applications, especially for high precision processing. The use of an ultra-short pulse laser is a promising alternative to the longer pulse lasers for machining thin and intricate materials for coronary stent applications. The use of femtosecond laser machining has been shown to reduce the size of HAZs, due to short beam/material interaction time. However, the main obstacle in applying this technology is the debris and recast formation, where the surface of the ablated material is usually surrounded by redeposition of vaporized material that has to be subsequently removed by other methods. Zheng et al. [116] performed studies of femtosecond laser processing nitinol sheet and used a nitrogen gas to reduce debris. Another alternative would be to remove it during the laser ablation process itself. This can be accomplished by laser machining under water for machining of silicon [153].

Laser processing in the presence of liquid water has been studied for more than 40 years for various applications [143]. Underwater laser processing provides a solution to some of the problems of laser machining in air and other assisting gases. Physical phenomena such as water convection and generation of bubbles aid the debris removal from the processed area. Kruusing [143] listed the advantages of water-assisted processing: light transmission, development of higher plasma pressure due to confinement, water carrying away debris, more effective cooling, useful chemical reactions, reduced pollution by waste gases and aerosols, reduced noise level and higher optical breakdown threshold than in air.
Chapter 7: Underwater femtosecond laser micromachining of thin nitinol tubes

The most common example of this method is water assisted silicon laser processing for various applications, e.g. micro-electro mechanical systems (MEMS), microfluidics, optics and photonics [154]. Water assisted laser ablation by using a femtosecond laser has been done in processing different materials under different experimental conditions [154-156]. Kaakkunen et al. [154] performed femtosecond laser drilling of silicon with the presence of a sprayed thin water layer onto the ablation spot during the drilling process. The ablation rate was increased and it was possible to obtain deeper and better quality holes than in ambient air. Machining in ambient air required twice the fluence and number of pulses compared to that with the presence of water. Water removes debris from the holes continuously, avoiding scattering caused by debris and resulting in a deeper hole with higher aspect ratio. Daminelli et al. [155] performed femtosecond laser irradiation of silicon under water confinement. They found that the debris redeposition was negligible and the ablated material remained suspended in the water layer. They observed, however, that ablation of silicon was more effective in air than under water [155]. It should be noted that there are materials (or experimental conditions) where the laser ablation rate in water is lower than in air and the opposite occurs under other conditions, as reported by Kruusing [157].

Results from studies of laser processing under water confinement are very promising, as the removal of the debris during the ablation process is beneficial, since the re-deposited debris and recast may scatter the laser beam and block the energy transfer on the material itself [154]. Underwater processing has been a deliberate approach to aid cutting to obtain better results. Yan et al. [158] reported underwater machining to deep cavities in alumina ceramic. They found that underwater machining has the capability of preventing the crack initiation, reducing heat damages and producing insignificant recast layer. Li and Achara [159] introduced a new technique of chemically assisted laser machining to minimize recast and heat affected zone, while increasing the material removal rate during drilling by using salt solution.
Underwater laser micromachining for tubes specifically for coronary stent applications received less attention. Work by Meng et al. [96] directed a water flow through the tubes during the fibre laser micromachining to reduce the heat affected zone (HAZ), as well as protecting the opposite surface of the tube. However, up to now, no work has been reported regarding the comparison of femtosecond laser micromachining of nitinol tubes in air and water environment, in particular for stent applications.

This chapter presents a technique for the application of water in laser micromachining compared to the air environment. The aim is to identify the beneficial effect of cutting in water, as well as to show that no assisting gas is required to remove the debris and minimize the recast, thus a significant manufacturing cost saving.

7.2 Experimental procedures

7.2.1 Experimental setup

A femtosecond laser machining of nitinol tubes was carried out by using a Coherent Libra Ti:Sapphire femtosecond laser that provides 800 nm wavelength, 100 fs pulse length, 1 mJ energy with a maximum average power of 1 W, 1 kHz pulse repetition rate and 100 mm focal length.

The femtosecond laser micromachining experimental setup is illustrated in Figure 7.1. A neutral density filter wheel was included in the optical train to enable the modification of the beam intensity. A power meter was used to measure the average power. The measurement was carried out before focusing the beam, where the power meter was placed just before the galvo head for an accurate reading of average power. A computer controlled galvo head scanner was employed to scan the beam in the X and Y direction and was controlled by a Wave Runner software.
Prior to the experiment, the kerf width was measured at different points to establish the focal plane. The minimal focal point was determined by the function of kerf width. The kerf width was found to reach the minimum value when the focal plane almost coincides with the top surface of the work piece. Zheng et al. [116] also suggested the minimum kerf width can be obtained when the laser beam was focused exactly on the surface of the sample. Figure 7.2 shows the effect of laser beam focal position, where the results show that at position 0 mm where the beam was focused on the surface of the sample it gave the minimum kerf width. Fixed focal position (0 mm) was used throughout the entire experiment for both cutting environment. The spot size at the focus was found to be 100 µm.
In this experiment, the frequency was fixed at 1 kHz. The experiments were performed by employing different fluences and cutting speeds. The fluence used was 4 J/cm$^2$, 6 J/cm$^2$, 8 J/cm$^2$ and 13 J/cm$^2$ (maximum fluence). At each fluence, the beam scanning speed was incremented from 0.1 mm/s to 0.5 mm/s, 1.0 mm/s and 1.5 mm/s. It is should be noted that the machining process with femtosecond laser required multiple-pass scanning to achieve full-depth penetration cut. In this experiment, different combinations of process parameters significantly required different number of passes which will be explained in section 7.3.1.

By using the aforementioned parameters setup, the cutting experiments were carried out in both air and underwater environments. In the initial arrangement, the tube was set horizontally on the stage with the cutting performed in air environment. In the second arrangement (underwater), the tube was placed horizontally inside an open petri dish. The petri dish was filled with a thin water film (1 mm thickness above the tube) at room temperature (298 K for pure water). The laser power reaching the work surface in water is attenuated following the Beer-Lambert’s law [143]:

$$I_x = I_0 \exp \frac{-x}{\Delta}$$  \hspace{1cm} (7.1)
Chapter 7: Underwater femtosecond laser micromachining of thin nitinol tubes

where $I_0$ is the entrance light intensity and $I_x$ is the light intensity at distance $x$, and $\Delta$ is the light absorption length. For pure water at 298 K, the absorption length for $\lambda = 800$ nm is 48 mm [143, 160]. In this work, 1 mm water level above the sample was used, with only 2% water absorption, allowing 98% of laser power reaching the target.

The specimen used in this experiment was nitinol tube, a shape memory metal alloy, where the two elements (titanium and nickel) are present in roughly equal atomic percentages. The tube has an outer diameter of 3.5 mm and 180 µm wall thickness. Nitinol is an established material used in self expanding stents due to unique properties: shape memory and super-elasticity. Nitinol material properties are listed in Table 7.1. The arrangement used for femtosecond laser micromachining of nitinol for air (dry machining) and underwater machining are shown in Figure 7.3.

<table>
<thead>
<tr>
<th>Properties</th>
<th>Values</th>
<th>Unit</th>
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<tbody>
<tr>
<td>Fusion point</td>
<td>1250</td>
<td>°C</td>
</tr>
<tr>
<td>Fusion heat</td>
<td>2522</td>
<td>J/cm³</td>
</tr>
<tr>
<td>Density</td>
<td>6.65</td>
<td>g/cm³</td>
</tr>
<tr>
<td>Thermal conductivity</td>
<td>10⁻¹⁸</td>
<td>W/mK</td>
</tr>
<tr>
<td>Thermal diffusivity</td>
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<td>cm²/s</td>
</tr>
<tr>
<td>Specific heat</td>
<td>490</td>
<td>J/kgK</td>
</tr>
<tr>
<td>Corrosion resistance</td>
<td>Very good</td>
<td></td>
</tr>
<tr>
<td>Biocompatibility</td>
<td>Very good</td>
<td></td>
</tr>
</tbody>
</table>

Table 7.1: Material properties of nitinol used in this work [39, 147].
The characterization of the laser cut samples was conducted by scanning electron microscopy (Hitachi S-3400N) for high resolution imaging. Keyence 3D Microscope was utilised to measure the kerf width. Topographical details of the cut edges were characterised by a MicroXam white light interferometer surface profile measurement system. The hardness values were measured using a MTS Nano-indenter XP (Agilent) to observe the heat affected zone (HAZ).
7.3 Results

7.3.1 Number of passes to achieve full-depth penetration cut

Multiple passes scanning were required to achieve full-depth penetration cut in femtosecond laser cutting. In this work, the ablation depth per pass was found to be within 1 to 15 µm depending on the process parameters and cutting conditions. It was also observed that the ablation efficiency was reduced after a certain number of passes due to debris from the previous scan blocking the laser beam path for the next scan. In this case, more passes were needed to achieve a complete cut. More number of passes were required in underwater cutting compared to that in cutting under air medium. Table 7.2 provides the number of passes required in dry and underwater cutting. It shows how this varies with laser fluence and cutting speed.

Table 7.2: Number of passes required to achieve full-depth penetration cut in dry and underwater

<table>
<thead>
<tr>
<th>Fluence (J/cm²)</th>
<th>Cutting speed (mm/s)</th>
<th>Number of passes</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>Dry</td>
</tr>
<tr>
<td>13</td>
<td>0.1</td>
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<td>38</td>
</tr>
<tr>
<td></td>
<td>0.5</td>
<td>65</td>
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</tbody>
</table>
7.3.2 The effect of the process parameters on the cut kerf width

Figure 7.4 shows the kerf width as a function of fluence (energy density) at different cutting speeds. According to the figure, kerf width was found to be approximately proportional to the fluence. The kerf width reduces as the cutting speed increases from 0.1 mm/s to 1.5 mm/s. In both cutting conditions, minimum kerf width was obtained at the lowest fluence and highest speed, giving the minimum kerf width of 80 µm. Based on the overall results, underwater machining produced kerf widths relatively larger than in dry condition. Similar observations were found by Tangwarodomnuukun et al. [161] in dry and underwater micromachining of silicon substrates. This phenomenon could be explained as the ability of pulsed laser ablation in liquid to create a high density shock-wave due to the liquid confinement effect, as discussed by Fabbro et al. [162]. The generation of vapour bubbles during the process introduced a significant impact force towards the workpiece, resulting in a wider kerf [161]. Findings by Fabbro et al. [162] also show that the impact pressure acting towards the workpiece is about 10 times greater than in the ambient air, which introduced mechanical assisted ablation [161, 162]. On the other hand, the defocusing effect by the water to the beam was one of the reason widening of the kerf.

Even though pulsed laser ablation in water generated impact forces during the process, the ablation depth for single pulse in underwater micromachining was found to be much lower than that in the dry condition. The laser intensity was weakened due to absorption by water, bubbles and debris in the water layer. Even with the impact force acting to push and carry the debris away from the keyhole, underwater ablation required nearly twice the number of passes compared with that in dry cutting to perform full depth penetration cutting at the same laser parameters setup.
Figure 7.4: Kerf width as a function of fluence at different cutting speeds for dry and underwater machining: (a) 0.1 mm/s (b) 0.5 mm/s (c) 1.0 mm/s (d) 1.5 mm/s.

### 7.3.3 The effect of the process parameters on the cut surface quality

Figure 7.5 shows the effect of laser fluence on the surface roughness, $R_a$, at different cutting speeds. In dry cutting, it can be seen that the surface roughness increased with the increasing fluence, and decreased with increasing cutting speed. At high fluences (13 J/cm$^2$ and 8 J/cm$^2$), the surface roughness was relatively high due to the presence of debris, and the formation of recast layers on the cut edge. When the fluence was reduced to (6 J/cm$^2$ and 4 J/cm$^2$), a smooth cut edge was obtained with significantly reduced recast. The high surface quality was obtained at low fluence and high cutting speed (4 J/cm$^2$, 1.5 mm/s), where the surface roughness was found to be minimal with $R_a = 0.29$ µm. Figure 7.6 shows the SEM images of the cut edges.
at a cutting speed of 1.5 mm/s, where the surface roughness was found to be minimal at all the fluence values. The figure also illustrates how the energy density affected the cut surface quality, as the surface improved when the fluence was reduced.

Figure 7.5: Surface roughness, $R_a$ as a function of fluence at different cutting speeds for dry and underwater machining: (a) 0.1 mm/s (b) 0.5 mm/s (c) 1.0 mm/s and (d) 1.5 mm/s.
Figure 7.6: Surface tomography at different fluences and the same cutting speed (1.5 mm/s) in dry machining: (a) 4 J/cm$^2$, (b) 6 J/cm$^2$, (c) 8 J/cm$^2$, and (d) 13 J/cm$^2$.

The opposite trend was observed in the underwater cutting, where the surface tended to be smoother with the increasing fluence and decreasing speed. The advantage of the underwater approach was that the surface produced was clean from debris and recast, as compared to that in the dry machining. This is due to the hydrodynamics of water over the workpiece surface, since the thin and thermally disturbed water layer prevents the re-deposition and the recast laser of melted nitinol and debris by removing them away from the cut region. However, it is important to note that an irregular surface (due to striations) was formed with underwater machining. The striations were more pronounced with decreasing fluence and increasing speed. The cut surface quality was degraded due to the formation of the striations, which created non-uniformity on the cut surface. Slow drifts and disturbances of water may be one of the reasons why the surface was corrugated. Work by Han et al. [163] also found this behaviour in processing silicon in water confinement. They explained that the corrugated surface obtained with underwater machining owing to that H$_2$O molecules
are strongly polarised due to the high electro-negativity of the oxygen atoms. During the process, the laser beam can be considered as a moving electromagnetic field on the sample surface. This moving electromagnetic field will affect the energy status of H$_2$O molecules because of its polarisation which influences the electromagnetic field itself in return. They also suggested that the surface quality may be improved at an appropriate laser pulse overlapping (with increased repetition rate) [163]. The high surface quality was indeed obtained at high fluence and low cutting speed (13 J/cm$^2$, 0.1 mm/s) where the surface roughness was found to be R$_a$ = 0.36 µm. Figure 7.7 shows the SEM images of the cut edges at a cutting speed of 0.1 mm/s, where the surface roughness was found to be minimal at all the fluence values. The figure illustrates how the energy density affected the cut surface quality as the surface improved when the fluence increased.

![Figure 7.7: Surface tomography at different fluences at the optimal speed, 0.1 mm/s in underwater machining: (a) 4 J/cm$^2$, (b) 6 J/cm$^2$, (c) 8 J/cm$^2$ and (d) 13 J/cm$^2$.](image)

Minimum surface roughness was obtainable in both dry and underwater approaches with R$_a$ = 0.29 µm and R$_a$ = 0.36 µm respectively. Dry cutting required low fluence and high speed (4 J/cm$^2$, 1.5 mm/s) while the opposite parameter setup at high
fluence and low speed (13 J/cm$^2$, 0.1 mm/s) was required by cutting in underwater to achieve this comparable quality. Detailed characterisation was carried out to observe the striations along the cut edges as shown in Figure 7.8 and 7.9 respectively. Both samples exhibited a shallow and uniform striations pattern, with the striation depth found to be around 0.7 µm for both samples.

Figure 7.8: Nitinol cut surface tomography for dry machining at 4 J/cm$^2$ and 1.5 mm/s: (a) cut edge (b) interferometer image giving the surface profiles data and (c) striations profile across the cut edge.
7.3.4 Heat Affected Zone (HAZ) analysis

The HAZ was assessed by performing the nano indentation tests to measure the material hardness variation. Each sample with different fluences at the cutting speed 0.1 mm/s for both cutting conditions was chosen for the test. The results obtained from the test showed a remarkable achievement with the femtosecond laser in regards of HAZ. Figure 7.10 shows the HAZ relationship with fluence at 0.1 mm/s cutting speed. HAZ was detectable (around 2 µm) for dry cutting at the highest fluence (13 J/cm²) with the hardness increased to 6.05 GPa as compared to the parent material hardness, 3.7 GPa. No presence of HAZ was observed when the fluence was reduced, with constant hardness observed adjacent to the cut edges (3.6 – 3.9 GPa). A remarkable achievement was observed with underwater cutting, as the cut samples...
were totally free from HAZ, with no significant changes to the material hardness observed (3.6 – 3.9 GPa). No changes of hardness indicate that no significant recrystallisation took place during the process.

![Graph showing HAZ at different fluences](image)

**Figure 7.10: HAZ at different fluences (cutting speed 0.1 mm/s).**

In the femtosecond laser pulse time period, the material removal process was dominated by material vaporisation. This involves rapid energy deposition to the material which causes the material to undergo a rapid temperature rise to the vaporisation state. It is expected that femtosecond laser will produce HAZ free results, due to the ultra-fast beam-material interaction time. In this work, the 2 µm HAZ was observed at 13 J/cm² in dry cutting. This could be explained as being due to the high thermal load and a longer period of vapour/plasma formation that reheats the surface causing the formation of a recast layer transferring the excessive heat to the bulk material. Figure 7.11 shows the formation of recast layer with dry cutting while underwater cutting shows a recast free cut on the cut edge even at the highest fluence (energy density).
Debris and recast formation

From the overall observation, machining with femtosecond laser resulted the accumulation of debris. Debris and recast formation were observed on each samples cut by in air environment. Figure 7.12 shows the debris accumulated on the perimeter of the cut kerf for nearly 0.8 mm. Apart from debris, prominent recast was also observed deposited around the edges of the ablation site. Conversely, underwater experiments produced cutting with free debris and recast (Figure 7.13). From the experiment observation, the colour of the water changed to greyish (showing that the ablated materials remained suspended in water layer phase), which showed that the debris was transported by the water during the process.
Figure 7.13: Underwater cutting with debris free and recast free result.

7.3.5.1 Ultrasonic cleaning of debris and recast for dry cutting

In section 7.3.2 the optimum surface quality were presented for both conditions, where $R_a = 0.29 \, \mu m$ for dry and $R_a = 0.36 \, \mu m$ for underwater cutting have been achieved. Clearly, both approaches are capable of producing fine surface finishes, with no detectable HAZ at the optimum parameter setup. It has been found that the main obstacle seen for dry cutting was the debris and recast formation as compared to underwater cutting. Subsequently, we performed the 15 minutes ultrasonic cleaning on the dry machined sample ($R_a = 0.29 \, \mu m$) to remove the debris and recast. The purpose is to compare the high cut surface quality dry machined sample after the ultrasonic cleaning with the one obtained in underwater cutting ($R_a = 0.36 \, \mu m$) in terms of debris and recast. The accumulated debris was easily removed by ultrasonic cleaning since the cotton-like debris was not permanently attached to the parent material. However, the recast layer, which is attached on top of the kerf edge was not able to be removed by using this method due to the strong adhesion to the parent material as shown in Figure 7.14. Although high cut quality was achieved with this parameter, debris and recast was prominent and required subsequent processing using chemical etching to clean the recast. Figure 7.15 shows the underwater machining sample free from debris and recast without any post processing.
7.4 Discussions

7.4.1 Debris and recast formation mechanism in dry cutting

Based on the results presented in Section 7.3, it has been seen that femtosecond laser are able to produce the high cut surface quality with no HAZ at the optimum parameter in both cutting environment. However, it has been found that the
drawback in applying femtosecond technology is the debris formation with dry cutting. Normally, multiple passes are required to cut through the samples and the process results in deposition of debris. Debris formation is a result of the condensation of supercooled vapours, partially on the sample surface, partially in the gas phase via collision with ambient gaseous molecules. The majority of the vapour initially ejected eventually finds itself back on the perimeter of the ablation pattern surface and inside the kerf as explained by Mai [164]. The travel velocity is in the order of several kilometres per second [165]. The debris formation degrades the feature quality and functionality of the components. It also reduces ablation efficiency, since debris from the previous scan can block the laser beam path for the next scan [164]. Figure 7.16 shows the schematic diagram of the interaction between fs laser beam and nitinol in dry machining.

![Figure 7.16: Schematic diagram of interaction between fs laser beam and nitinol in dry machining.](image)

Large numbers of previous publications have suggested the use of underwater machining approach to remove debris and recast during machining [154-157, 161].
The underwater laser micromachining process is more complex than laser micromachining in ambient air since there are several other phenomena and effects involved in this process. The interaction between femtosecond laser and nitinol through water led to a generation of a plasma in the water medium through a multiphoton ionization process [166]. Excited electrons and plasma transfer their energies to the nitinol lattice and the water molecules, hence creating a rapid rise in temperature that generates stress in the water and induce the emission of a strong pressure wave that propagates in the water. The energy transfer is strong enough to induce a phase transition in the water, thus creating a cavitation bubble. It is expected that the generated temperatures reached ~10 000 K and pressures can be in the range of 2~7 GPa [153, 167]. Plasma collisions and relaxation has been shown to be the main source of mechanical stress in the medium and the dominant factor for the pressure wave and bubble creation [168]. The high recoil pressure by water evaporation would benefit the prevention of recast and debris formation. It is also known that phenomena such as supercontinuum generation, filamentation and optical breakdown of the water phase have a shielding effect, which reduces the amount of energy reaching the target. For a pulse duration of 100 fs, plasma transmission in water decreased from 50% to 20% as reported by Daminelli et al. [155]. Figure 7.17 illustrates the interaction between the femtosecond laser beam, water and nitinol during the underwater machining.
On the other hand, underwater phenomena such as bubble creation played important role in the process as it affect the surface quality. The corrugated effect is more prominent when the cutting was performed at the high speed (1.5 mm/s) as high speed required more passes to complete the cutting subsequently increase the time for the bubble reaction with the target. Figure 7.18 shows the example of corrugated surface generated by underwater cutting at 13 J/cm² and 8 J/cm² at 1.5 mm/s which required high number of passes, 125 and 150 passes respectively to obtain the complete cut. This effect can be avoided by employing a thin flowing water film during the machining process. With the water flowing at the high flow rates (suggested flow rates greater than 1 m/s), this will reduce the localised bubble effect at the ablated surface as compared to the confined water layer [164].
7.4.2 Approach in preventing recast and debris formation

Other than underwater cutting which is known as effective approach in preventing debris and recast formation, the cutting process was carried out with the aid of compressed air (pressure = 2 bar) along the sides of the cutting direction to minimise debris and recast formation. The result obtained showed that the debris and recast formation was similar with obtained in ambient air as shown in Figure 7.19.

Figure 7.19: (a) Debris and recast formation during the cutting process assisting by compressed air (b) close up view.

Generation of debris and recast layer with femtosecond laser machining has generated interests among researchers seeking a method to reduce this problem.
Zheng et al. [116] employed a stream of N\textsubscript{2} gas during the femtosecond laser cutting of nitinol sheet to assist in reducing the amount of the redeposition material. However, redeposition in powder form along the cut was observed. Work by Matsumura et al. [169] compared femtosecond laser machining silicon substrate in gas environment (Ar, He and N\textsubscript{2}) and in vacuum. They reported that the machining process also produced significant amount of debris in the gas environment whereas no debris were found in vacuum. Mai [164] also reported that a cleaner machining surface was observed under vacuum. He also reported another technique by depositing water soluble coating, commonly polyvinyl alcohol (PVA) to trap debris particle. Choo et al. [153] introduced an alternate method for trapping the debris generated on the finished surface. The method involved covering the target material by using an adhesive tape resulting in absence of debris and redeposited of molten material on the surface after the tape was removed. Low et al. [170] used a ceramic powder imbedded polymer coating to prevent spatter formation during laser drilling of aerospace materials.

In the current study, the test material is a small diameter tubing but quality of the back wall must also be put into consideration. The findings from this study suggest that water assisted cutting is the most effective method to achieve high machining quality with debris-free and recast-free cutting, as well as a clean back wall as shown in Figure 7.20 compared to that in air. In general, the presence of water minimises the re-deposition of ablated material, resulting in better cutting quality of surface finish and eliminating the need to remove the debris generated subsequently. In addition, it was observed that the use of water could reduce the emission/liberation of waste gases and particles into the atmosphere. High machining quality with well defined edges, debris-free and recast-free was achieved straight away after the cutting process has shown that underwater cutting can reduce cost to the manufacturing process by reduce the need for post-process cleaning.
7.5 Conclusions

The effects of femtosecond laser micromachining of nitinol in air and underwater have been investigated. Both quantitative and qualitative differences were reported. In dry cutting, minimum surface roughness was obtained at the low fluence (4 J/cm$^2$) and high speed (1.5 mm/s). The debris was removable with ultrasonic cleaning whereas the recast was permanently attached to the material requiring subsequent processing to be removed. 2 µm of HAZ was observed at the highest fluence (13 J/cm$^2$) and no presence of HAZ was observed when the fluence was reduced. Conversely, in underwater cutting, high fluence (13 J/cm$^2$) and low speed (0.1 mm/s) was used to obtain the high surface quality. No presence of HAZ was observed for underwater cutting, which was debris-free and recast free with clean opposite back walls. It was observed that both dry and underwater cutting conditions gave high surface quality at the optimum parameters setup. However, in the dry cutting, the debris and recast degrade the cut quality produced. Large kerf width was obtained in this work due to large beam spot size (100 µm), owing to the limitations of the optical setup.
Chapter 8

Understanding the behaviour of pulsed laser dry and wet micromachining processes by multi-phase smoothed particle hydrodynamics (SPH) modelling

A Smoothed Particle Hydrodynamics (SPH) numerical model is developed to simulate the three-phase laser micro-machining process for medical coronary stent manufacture. The open-source code SPHysics is used to model the interaction between the laser beam and workpiece. This enables the melt flow behaviour in the non-linear pulsed fibre laser micro-machining process to be modelled. The developed model considers the conversion of laser energy into heat within a very thin surface layer, heat conduction into the parent material and the phase transition between solid, liquid and vapour. A good agreement with experimental data is obtained for predicting the penetration depth and melt ejection velocity. Water is also incorporated in this model to help explain the mechanism in laser wet micro-machining and drilling. It is
demonstrated that the meshless characteristics of SPH are able to model the droplets ejected from kerf where it is difficult for conventional modelling. A static beam was used throughout the model development.

8.1 Introduction

Laser micro-machining of metallic materials is mainly a thermal process where the material is removed due to melting, vaporisation and the action of assist gas unless the laser used is in the femtosecond pulse range. During the process, the work piece undergoes phase changes from solid, liquid and vapour which involve significant changes of material properties. With the aim of understanding the mechanisms and improving the process itself, numerous modelling methods have been used for modelling the laser cutting process including analytical [171], Finite Element (FE) [172, 173], Finite Difference (FD) [174] and Finite Volume (FV) [175]. Conventional modelling methods using generated mesh have been problematic in capturing the melt splashing or droplets during the material removal. The meshes need to be several times smaller than the splashing of ejected molten materials, which requires massive computing resources. It is even more problematic when modelling micro cutting where the ejected droplets can form at scales as small as micro/nano scale [176]. These established methods have succeeded in modelling laser cutting but without sufficient accuracy to be extended to micro-machining where the material removal mechanism is different. For example, with finite element simulation, the modelling work in laser micromachining reported by Shalahim et al. [172] included the phase changes in the model based on the temperature distributions. However, the behaviour of the ejected particles is not able to mimic the physical process due to limitation of this method.

The smoothed particle hydrodynamics (SPH) method is foreseen as being able to capture these nonlinear phenomena. SPH is a meshless computational method represented by a set of particles where each particle moves according to the governing dynamics. In SPH, differential equations are therefore solved as a Lagrangian
technique. The advantages of this method compared to other computational methods are due to; (i) its meshless nature such that no mesh or grid is required to compute spatial derivatives, and (ii) the Lagrangian approach can capture nonlinear effects inherently. This method was initially formulated to simulate non-axisymmetric phenomena in astrophysics proposed by Lucy [177] and Gingold et al. [178] in 1977. In recent years, the flexibility of this method suited in various research fields has been increasingly used in astrophysics [178, 179], oceanography [180-186], volcanology [187] and solid mechanics [188, 189].

Work by Zhang et al. [190] investigated droplet spreading and solidification with SPH followed by Xiong and Zhu [191] in modelling thermal sprays to demonstrate the applicability of SPH to handle such problems. This is similar to the melt and solidification processes involved in laser cutting. The phenomenon where the splashing of melt ejection leaves the kerf has similarity with the problems in water waves in hydrodynamics which is successfully modelled by using SPH presented by Dalrymple and Rogers [180].

The applications of SPH in the field of laser processing are still in the initial stages and have not yet been developed. To date, only a few publications have been found in this area. These include the thermal modelling of direct laser interference patterning by Demuth et al. [192], modelling of the heat transfer and fluid flow during laser spot welding by Tong et al. [193], the effect of hydrodynamics of water on debris removal during laser underwater machining of ceramics by Yan et al. [158] and modelling of laser cutting process [176] by Gross. Laser processing is a new modelling field to be explored by using SPH to obtain a reasonable accuracy. The first work in SPH modelling of laser cutting has been reported by Gross [176]. He outlined the basic foundations and has demonstrated the early development of SPH modelling for application in laser cutting. The model included the phase changes during laser cutting process; however, the results are at early stages of demonstrating the process.

There is currently no established model for the simulation of laser micro-machining of metallic materials by SPH and this work aims to fill this knowledge gap. The SPH
method was chosen due to the meshless characteristics which have been demonstrated in a wide range of applications. In this chapter, the SPH method is introduced for the first time in laser micro-cutting and micro-drilling applications.

This chapter describes the physics behind the laser micro-machining process that is solved with a three-dimensional SPH model. Here in, a static beam was used throughout the simulation to model the beam-material interaction. First, the results for full-depth penetration and melt ejection velocity are presented. The formation of recast and spatter is also presented. Second, the simulation is performed with the presence of water underneath the workpiece, to help explain the hydrodynamic effects in wet machining to minimise backwall damage.

8.2 Model description

8.2.1 Physical process to be captured in SPH model

The laser micro-machining and micro-drilling process considered in the model is pulsed mode fibre laser machining of thin stainless steel 316L. The power profile of the laser beam is a Gaussian distribution and the pulse duration considered here is in millisecond regime. Nitrogen was utilised as an assist gas. The configuration of the laser and the workpiece is shown in Figure 8.1. Initially, the laser beam is projected onto the top surface. The workpiece is then rapidly heated, melted and vaporised [18, 194]. The molten liquid is ejected from the cut kerf due to pressure exerted by assist gas (N\textsubscript{2}) along with recoil pressure produced by surface vaporisation [194]. N\textsubscript{2} as an assist gas enhances the melt ejection mechanism by adding more pressure to the recoil pressure acting as mechanical force to remove the molten material.
The physical processes to be captured by SPH modelling in this work are illustrated in Figure 8.2 for a clearer depiction. The figure provides the sequence of the cutting/drilling process until the full penetration depth is achieved with a static beam. The initial beam-workpiece arrangement is shown in Figure 8.2a before the drilling process begins. Figure 8.2b shows the process when the beam has heated the target and partial penetration depth is achieved. The material is ejected upward due to the effect both of assist gas and recoil pressure where the heat propagates throughout the material until the full-depth penetration is achieved as shown in Figure 8.3a. When it reaches the full depth penetration, the materials due to the melt ejection exit both upward and downwards out of the kerf. Recast and spatter also accumulated on the cut surface.

Water can also be included in the model to study the hydrodynamic effect in wet machining to minimise backwall damage. In the experiments, water is introduced inside the internal diameter of the tube. In the dry machining, once full-depth penetration is achieved, the ejected materials are found to adhere to the back wall resulting in significant back wall damage. With wet machining, the ejected molten particles are carried away by the flowing water resulting in clean back wall damage as illustrated in Figure 8.3b.
8.2.2 Experimental procedures for verification

A GSI JK100FL single mode fibre laser was utilised in the experiment. The fibre laser has a 1080 ± 5 nm output wavelength, 25 μm theoretical spot size at the focal plane and a beam quality factor, $M^2 < 1.1$. The coaxial assist gas nozzle had an exit diameter of
0.5 mm. The thin stainless steel 316L workpiece with 150 µm thickness is penetrated with the laser by using single pulse. Nitrogen, N\textsubscript{2} was used as an assist gas. The experiments were performed with the following laser parameters: Peak power, $P_\text{p}$ = 100 W, pulse duration, $\tau$ = 0.05 ms, $\tau$ = 0.10 ms and $\tau$ = 0.15 ms and N\textsubscript{2} gas pressure of 6 bar.

### 8.2.3 Physical Properties

The thermophysical properties of the stainless steel 316L and N\textsubscript{2} assist gas used in the modelling are shown in Tables 8.1 and 8.2. The properties of water are also included in Table 8.3.

#### Table 8.1. Thermophysical properties of stainless steel 316L [195, 196].

<table>
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<th>Value</th>
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<td>Density, $\rho$ (kg m\textsuperscript{-3})</td>
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<tr>
<td>Specific heat, $c$ (J kg\textsuperscript{-1}K\textsuperscript{-1})</td>
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<tr>
<td>Thermal conductivity, $k$ (W m\textsuperscript{-1}K\textsuperscript{-1})</td>
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<tr>
<td>Latent heat of fusion, $L_v$ (J kg\textsuperscript{-1})</td>
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<td>Initial temperature, $T_0$ (K)</td>
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<td>Melting temperature, $T_m$ (K)</td>
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<td>Boiling temperature, $T_b$ (K)</td>
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#### Table 8.2: N\textsubscript{2} assist gas and nozzle parameters [197].

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<th>$\text{N}_2$ properties and nozzle parameters</th>
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<tr>
<td>$\text{N}_2$ specific heat ratio, $\gamma$</td>
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</tr>
<tr>
<td>Assist gas pressure, $P_i$ (Pa)</td>
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</tr>
<tr>
<td>Nozzle exit diameters, $d_n$ (m\textsuperscript{2})</td>
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</tr>
<tr>
<td>Nozzle workpiece distance, $z_n$ (m)</td>
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</tr>
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</table>

#### Table 8.3. Properties of water [198].

<table>
<thead>
<tr>
<th>Properties of water</th>
<th>Value</th>
</tr>
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<tbody>
<tr>
<td>Density, $\rho$ (kg m\textsuperscript{-3})</td>
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</tr>
<tr>
<td>Thermal conductivity, $k$ (W m\textsuperscript{-1}K\textsuperscript{-1})</td>
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</tr>
</tbody>
</table>
8.3 Mathematical formulation

8.3.1 Model assumptions

To develop a numerical model of the dry and wet laser machining process, the following assumptions were introduced in building the model:

- Laser beam was Gaussian in energy distribution across the beam.
- Material was isotropic with constant thermal and optical properties.
- Plasma generation was ignored. Only solid, molten liquid and vapour phase were considered [194].
- No interaction between laser beam and the vapour. This is because vapour is optically thin (transparent), and therefore does not absorb much energy from laser beam [194, 199].
- The generation of shock waves is ignored.
- Zero assist gas exists outside the beam area [194].
- No gas particles were created in the simulation. The assist gas included is for the purpose of ejecting the molten particles only.

8.3.2 Heat transfer

In this study, the governing equation of heat transfer is expressed as [200]:

\[
\frac{c_p}{\rho} \frac{dT}{dt} = \nabla \cdot (k \nabla T) + Q - (Q_v)
\]  

(8.1)

where \( T \) is the absolute temperature, \( c_p \) is the heat capacity per unit mass at constant pressure, \( \rho \) is the density, \( k \) is the coefficient of thermal conductivity, \( Q \) is the laser source term and \( Q_v \) is the heat loss.

Heat loss is mainly due to convection and radiation, where \( Q_v \) is given by [201],

\[
Q_v = h_c(T_s - T_0) + \varepsilon \sigma (T_s^4 - T_0^4)
\]  

(8.2)
where \( h_c \) is the heat convection factor = 20 W m\(^{-2}\)K\(^{-1}\), \( \varepsilon \) is emissivity (0.4 is used for SS 316L), \( \sigma \) is Stefan-Boltzmann coefficient = 5.67 x 10\(^{-8}\) W m\(^{-2}\)K\(^{-4}\), \( T_s \) is the surface melting temperature and \( T_0 \) is the initial temperature [201].

The Gaussian beam distribution can be described as follows [201]:

\[
Q = \frac{P_{\text{laser}}}{r_b^2 \pi} \exp\left(-\frac{2r^2}{r_b^2}\right) \tag{8.3}
\]

where \( P_{\text{laser}} \) is the laser power, \( r \) is the radial distance from the beam centre and \( r_b \) is the laser beam radius.

### 8.3.3 Melt ejection velocity

Melt ejection velocity, \( V_m \), is required in the model for ejecting molten particles. It combines the effect of the assist gas pressure and vapour pressure which can be determined from Bernoulli’s equation which was simplified to be [194]:

\[
V_m = \sqrt{\frac{2(P_{\text{vap}}+P_{\text{eff}})}{\rho_m}} \tag{8.4}
\]

where \( P_{\text{vap}}, P_{\text{eff}}, \rho_m \) are vapour pressure, effective assist gas pressure and melt density, respectively.

### 8.3.4 Vapour pressure

Vapour pressure, \( P_{\text{vap}} \), was determined by using the Clausius-Clapeyron equation [194]:

\[
P_{\text{vap}} = P_0 \exp\left[\frac{L_v}{R} \left(\frac{1}{T_b} - \frac{1}{T_s}\right)\right] \tag{8.5}
\]

where \( P_0, L_v, R, T_b \) and \( T_s \) are atmospheric pressure, latent heat of vaporisation, the gas constant, boiling temperature and melt surface temperature, respectively.
8.3.5 Effective assist gas pressure

In gas-assisted cutting, the assist gas pressure helps the acceleration of ejected materials rather than depending solely on the recoil pressure. Here, we include the effect of the assist gas pressure induced on the melt surface in the model.

At the nozzle exit, the assist gas experiences adiabatic expansion and thus accelerates up to the local speed of sound which leads to the critical state [194, 202, 203]. Thus, the critical assist gas, $P_c$ can be calculated from:

$$P_c = \left( \frac{2}{\gamma+1} \right)^{\gamma-1} P_i$$

(8.6)

where $P_i$ is the pressure inside the nozzle and $\gamma$ is assist gas specific heat ratio. For diatomic gases such as nitrogen, $\gamma = 1.4$ [197]. Assist gas pressure acting in the kerf entrance is known as effective assist gas pressure, $P_{eff}$ can be defined as [194]:

$$P_{eff} = P_c \frac{A_{eff}}{A_{eff} + A_{rl}}$$

(8.7)

where $A_{eff}$ is the effective area of flow entering the hole defined by the laser beam radius, $r_1$. $A_{rl}$ is the cylindrical area where the radial loss of the gas pressure flows as defined by the nozzle exit diameter, $d_n$ and nozzle workpiece distance, $z_n$ as follows:

$$A_{eff} = \pi r_1^2$$

(8.8)

$$A_{rl} = d_n \pi z_n$$

(8.9)

8.3.6 Modelling the phase changes

During the laser micro-machining process, if the energy of the laser beam incident with the solid material is sufficient, melting starts and a solid-liquid interface is formed and propagates into the solid. The phase change of the parent material is simulated with a temperature-density dependent relationship, where the density of liquid state, $\rho_m$ is given by Equation 8.10 [195]:

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Chapter 8: Understanding the behaviour of pulsed laser dry and wet micromachining processes by multi-phase SPH
\[ \rho_m = 6881 - 0.77(T_s - 1723) \]  

(8.10)

where \( T_s \) is the melt surface temperature.

### 8.4 SPH Model – SPHysics

#### 8.4.1 Governing equations

The work presented herein was developed by using the free open-source SPH code known as SPHysics [204]. The SPH method is expressed by a local interpolation for set of particles. These particles act as the interpolation points where the properties of the particles are calculated and updated for every time step. Each fluid particle is characterised by their properties; mass \( m_i \), density \( \rho_i \), pressure \( P_i \), velocity \( v_i \) and volume \( \omega_i \).

In SPH, the integral interpolated value of a function \( A \) at position \( \mathbf{r} \) is described in the following expression [204]:

\[
A(\mathbf{r}) = \int A(\mathbf{r'})W(\mathbf{r} - \mathbf{r'}, h)d\mathbf{r'}
\]

(8.11)

where \( h \) is the smoothing length and \( W(\mathbf{r} - \mathbf{r'}, h) \) is the kernel function. The integral interpolant is approximated by a summation interpolant at a particle \( i \)

\[
A(\mathbf{r}) = \sum_i m_j \frac{A_j}{\rho_j} W_{ij}
\]

(8.12)

where \( m_j \) and \( \rho_j \) are the mass and density, respectively, \( W_{ij} = W(|\mathbf{r}_i - \mathbf{r}_j|, h) \) is the kernel function with \( \mathbf{r}_{ij} = |\mathbf{r}_i - \mathbf{r}_j| \) is the distance between particles \( i \) and \( j \) and \( h \) is the smoothing length.

The smoothing length characterises the spatial extent of the kernel which is corresponding to an average cell size. In SPHysics, a smoothing length of \( h = 1.3\Delta \) is used where \( \Delta \) is the initial particle distance as this provides a good compromise between sufficient particles existing in the support of the kernel and accuracy. The
summation of Equation (8.12) refers to the particles \( j \) which are considered as the nearest neighbours to particle \( i \) contained within the kernel as illustrated in Figure 8.4.

![Figure 8.4: Neighbours of particles i within a support kernel.](image)

According to Morris et al. [205], in general a kernel function can be expressed as

\[
W(r_{ij}, h) = \frac{1}{\pi^d} f \left( \frac{r_{ij}}{h} \right)
\]

where \( d \) is the number of dimensions, \( h \) is the smoothing length and \( f \) is the function.

There are several kernel formulations available in the literature and the cubic spline kernel was chosen in developing the model. The kernel based on the spline functions has been found to be computationally efficient with compact support where the interactions are exactly zero for \( r > 2h \) [206]. The cubic spline kernel defined by Monaghan [206] is:

\[
W(r_{ij}, h) = \frac{1}{h^n} \left\{ \begin{array}{ll}
1 - \frac{3}{2} q^2 + \frac{3}{4} q^3 & 0 \leq q \leq 1 \\
\frac{1}{4} (2 - q)^2 & 1 \leq q \leq 2, \\
0 & q \geq 2
\end{array} \right.
\]

where \( q = \frac{r_{ij}}{h} \) and the term \( \frac{1}{h^n} = \frac{1}{n^3} \) for three dimension modelling.
The conservation of mass and the conservation of momentum expressed in SPH form are \[206\]:

\[
\frac{d\rho_i}{dt} = \sum_j m_j \mathbf{v}_{ij} \cdot \nabla_i W_{ij} \tag{8.15}
\]

\[
\frac{d\mathbf{v}_i}{dt} = -\sum_j m_j \left( \frac{\rho_i + \rho_j}{\rho_i \rho_j} \right) \mathbf{v}_{ij} W_{ij} \tag{8.16}
\]

where \( \mathbf{v}_{ij} = \mathbf{v}_i - \mathbf{v}_j \). It is should be noted that the momentum equation uses a form of the momentum equation derived from variation principles \[207\] since this performs better for multiphase simulations \[208\].

In this work, artificial viscosity, \( \Pi_{ij} \) proposed by Monaghan \[206\] is used to prevent the simulation from developing unphysical oscillations and stabilise the numerical algorithm \[209\]. The artificial viscosity \( \Pi_{ij} \) is added to the momentum equation as:

\[
\frac{d\mathbf{v}_i}{dt} = -\sum_j m_j \left( \frac{\rho_i + \rho_j}{\rho_i \rho_j} + \Pi_{ij} \right) \mathbf{v}_{ij} W_{ij} + \mathbf{g} \tag{8.17}
\]

where gravitational acceleration, \( \mathbf{g} = (0,0,-9.81) \text{ms}^{-2} \) and \( \Pi_{ij} \) is the viscosity term.

\[
\Pi_{ij} = \begin{cases} 
-\alpha \frac{c_{ij} \mu_{ij}}{\rho_{ij}^2} & \mathbf{v}_{ij} \cdot \mathbf{r}_{ij} > 0 \\
0 & \mathbf{v}_{ij} \cdot \mathbf{r}_{ij} < 0 
\end{cases} \tag{8.18}
\]

with

\[
\mu_{ij} = \frac{h v_{ij} r_{ij}}{r_{ij}^2 + \eta^2} \tag{8.19}
\]

\[
\bar{\rho}_{ij} = \frac{1}{2} (\rho_i + \rho_j) \tag{8.20}
\]

\[
\bar{c}_{ij} = \frac{1}{2} (c_i + c_j) \tag{8.21}
\]

where \( \eta^2 = 0.01 h^2 \), \( \alpha \) is a free parameter that can be changed according to the problem. In this work, the value of \( \alpha = 5 \) was used to make the solid phase behave as a
liquid with a very high viscosity stability in modelling the solid phase and $\alpha = 0.1$ was used for modelling the water.

To close the governing equations (continuity and momentum), the following equation of state was used to formulate the pressure as a function of density. The equation of state used widely in modelling weakly compressible fluid gives the pressure for each particle defined as the following [200]:

$$P = B \left[ \left( \frac{\rho}{\rho_0} \right)^\gamma - 1 \right]$$  \hspace{1cm} (8.22)

where $B = \frac{c_0^2 \rho_0}{\gamma}$, the reference density, $\rho_0 = 1000$ kg m$^{-3}$ and $c_0 = c(\rho_0) = \sqrt{\frac{\partial \rho}{\partial \rho}}$ and $\gamma = 7$.

We represent the continuity and momentum equations coupled with heat transfer in SPH form as:

$$c_{p,i} \frac{dT_i}{dt} = \sum_j \frac{m_{i,j}}{\rho_{i,j}} \left( k_i + k_j \right) \left( \frac{T_i - T_j}{r_{ij}} \right) r_{ij} \nabla_i W_{ij} + Q - (Q_v)$$  \hspace{1cm} (8.23)

where $W_{ij}$ is the smoothing kernel function and $r_{ij}$ is the magnitude of the distance between particles $i$ and $j$. Equations (8.15, 8.17 and 8.23) were integrated in time using a second-order predictor corrector scheme and a variable time step [210].

### 8.4.2 Boundary conditions

We have utilised dynamic boundary conditions where particles are arranged in a staggered position. The boundary particles were treated in the same way as fluid particles with the same governing equations (e.g. continuity equation, momentum equation and equation of state). The difference is that boundary particles are assigned to be fixed in space and model an impermeable wall to ensure the particles stay inside the domain. This has been chosen due to the computational simplicity [210].
8.4.3 Detection of surface particle

In the laser cutting process, the heat input from the laser source is active only on the open surface of the workpiece and melted particles are ejected once their temperature exceeds the melting point. To enable this, a method of surface detection was implemented for detecting the surface particles to activate the laser for those particles only, and allow the penetration of the laser beam through the material and at the same time enable the particle ejection.

The divergence of a particle position was implemented to help identify the surface particles. The divergence of a particle position in SPH is given by:

$$\nabla \cdot \mathbf{r} = \sum_j \frac{m_j}{\rho_j} \mathbf{r}_{ij} \cdot \nabla W_i(\mathbf{r}_{ij})$$

(8.24)

For the 3-D simulations performed here, an optimum criterion was found empirically such that $\nabla \cdot \mathbf{r} = 2.4$ used to determine the surface particles. This method has proven effective and adjustable depending of the applications and the compact support $h$ as illustrated in Figure 8.5 [211, 212].

![Figure 8.5: Zoom of a solid surface of a surface particle. The incomplete kernel support of a surface particle gives a non-zero value for divergence $\nabla \cdot \mathbf{r}$.](image)

8.4.4 SPH simulation procedure

In the simulation, the domain was represented by a set of particles that change over time to simulate the process. Two cases were investigated here known as dry and wet
machining. A three-dimensional domain was set up to be 100x200x150 µm$^3$ for dry cutting and 100x200x150 µm$^3$ for wet machining with 100 µm water thickness underneath the metal in $x$, $y$ and $z$, respectively. The particle size used was $\Delta x = 5 \times 10^{-6}$ m. A total of 35624 and 49760 particles were used in simulating dry and wet machining, respectively. The thickness of the stainless steel was set up to be 150 µm to agree with the experimental data. However a thinner water thickness was used to minimise the simulation time. The time step used was $\Delta t = 5 \times 10^{-8}$ s for both cases.

Figure 8.6: Neighbours of particles $i$ within a support kernel. Neighbours of particles $i$ within a support kernel. Neighbours of particles $i$ within a support kernel. Neighbours of particles $i$ within a support kernel. Neighbours of particles $i$ within a support kernel.

shows the geometry of the workpiece (parent materials) and Figure 8.6b represents the workpiece with the presence of water underneath. The laser peak power used is $P_p = 100$ W and laser beam radius is, $r_b = 25$ µm. The workpiece and water initial temperatures are assigned to be at room temperature, $T_0 = 300$K (assuming the laser processing takes place at room temperature).

Figure 8.6: Domain for SPH simulation (a) dry cutting (b) wet cutting.
8.5 Results and Discussions

8.5.1 Penetration depth

Full penetration depth is obtained when the depth of the laser penetration reaches through the entire extent of the material’s thickness, which in this case is 150 µm (SS 316L). This indicates that the materials are fully removed within the beam interaction zone down to the full material’s thickness. The modelling of penetration depth is significant in laser machining (drilling) to predict the full-depth penetration time for different laser parameters and different materials with different thickness.

Figure 8.7 assembles six snapshots from the SPH model showing the temperature profile within the beam active zone which indicates penetration depth at different time when firing with a single pulse laser (laser peak power, $P_p = 100$ W and pulse duration, $\tau = 0.15$ ms). The laser beam was placed in the middle of the workpiece, transferring the energy and heating the workpiece. The solid workpiece was heated at the initial temperature, $T_0 = 300$ K. The surface temperature increases and upon reaching the melting point, ($T_m = 1723$ K) the phase transition from solid to liquid occurs. The melt is expelled from the kerf and breaks up into droplets. Superheating of the liquid phase takes place in the irradiated region and reaching the boiling temperature ($T_b = 3100$ K), where the vaporisation state was reached and the ejected particles become optically thin and were assumed to not interact with the incident laser beam. The assist gas and the presence of the recoil pressure help to accelerate the material removal. The material in the vicinity to the beam interaction zone is then heated by heat conduction. It is demonstrated that the meshless characteristics of SPH are able to model the melt splashing and droplets leaving the kerf where it is found difficult for conventional modelling.
Figure 8.7: Cross section of the workpiece showing the phase changes in dry cutting. Full depth penetration with single pulse (Peak power, $P_p = 100$ W, pulse duration, $\tau = 0.15$ ms).

To validate the current model, the simulation results were compared against experimental data. Experimentally, three different pulse durations were used to obtain the penetration depth with 100 Watt laser peak power with single pulse firing. The
pulse durations used in three separate experiments were $\tau = 0.05$ ms, 0.1 ms and 0.15 ms where the latter achieved full depth penetration. The SPH simulation also was performed with the same laser parameters used in experimental work. Figure 8.8 shows a plot of the computed penetration depth with different pulse durations along with the experimental data. It was found that at $\tau = 0.05$ ms and 0.1 ms, respectively, the model slightly under and over predicts the penetration depth with $\sim 20$ $\mu$m differences than obtained experimentally. It can be seen that the model obtained full depth penetration at $\tau = 0.125$ ms which is less time required by the experiment. However, in the experiment we have only performed the process at the aforementioned pulse duration and did not study in a small time difference. The model suggests that in the physical process, the penetration depth might have occurred within this time which is known as the threshold time for full-depth penetration. The agreement between SPH and experiment is reasonable where the SPH model allows penetration depth prediction for a different laser power at single pulse. It is important to note that the experimental data set has some limitations for a comprehensive validation such as restricted pulse durations. Figure 8.9 displays a set of penetration depth images obtained experimentally at three different time instants corresponding to the experimental data plotted in Figure 8.8 for validation.

![Figure 8.8: Penetration depth evolution as a function of time obtained by the SPH model and experimentally at different pulse durations ($\tau = 0.05$ ms, $\tau = 0.1$ ms and $\tau = 0.15$ ms) with 100 W peak power at a single pulse.](image-url)
Chapter 8: Understanding the behaviour of pulsed laser dry and wet micromachining processes by multi-phase SPH

Figure 8.9: Penetration depth obtained experimentally at different pulse duration with 100 W peak power at single pulse (a) \( t = 0.05 \) ms, (b) \( t = 0.1 \) ms and (c) \( t = 0.15 \) ms.

### 8.5.2 Phase changes

As mentioned earlier in the introduction, laser cutting is a complex process involving phase changes, such as solidification, melting and evaporation. Other numerical methods experience difficulties in modelling the phase changes especially when it involves ejection melts breaking into droplets where a small mesh size is needed to be able to capture the droplet. Figure 8.10 shows a plot of the temperature as a function of time to obtain the point when the liquid and vapour phase initially formed during the process. It was found that the liquid phase was initially formed between time regime: \( t = 0.004 \) and \( t = 0.005 \) ms where the vapour initially formed at \( t = 0.011 \) and \( t = 0.012 \) ms with a small amount of surface particles reaching the boiling point. The melting and boiling temperatures are given as the thermal properties of the specimen.

Figure 8.10: Temperature plot indicate the phase change in the material obtained by SPH model obtained from the SPH model. \( (P_p = 100 \text{ W}, \ t = 0.15 \text{ ms}) \).
8.5.3 Melt ejection velocity

During the laser cutting process, material is removed from the kerf via ejection of molten material and also by vaporisation. Initially, when the workpiece receives the laser energy, the surface starts heating and later, the surface temperature can reach the melting point where the molten layer starts to form. The molten material is driven out along the wall from the cut kerf by the combined effect of the assist gas and the presence of recoil pressure. Soon after exiting the kerf, the molten material then breaks up into droplets and then starts to solidify [194, 213].

If the melt surface temperature significantly exceeds the melting temperature and the evaporation rate is low enough to produce noticeable recoil pressure, the melt removal is dominated by the effect of assist gas. Vaporisation recoil pressure becomes the primary factor of material removal if the melt surface temperature significantly increased. At the boiling surface temperature, melt removal due to evaporation takes place [214].

Figure 8.11a illustrates the average melt ejection velocity within the kerf against penetration depth. It was found that the particles were expelled from the kerf by both effect of assist gas and recoil pressure. It can be seen in region I, the particles ejection start with initial burst with low velocity, where a small amount of materials are ejected from the kerf. This is followed by a distance of 50 µm, where more materials experienced the energy causing more particles being ejected and the particle ejection velocity starts to increase where the effect of recoil pressure becomes significant.

In region II (half penetration depth), the ejection velocity of melt particles is even higher. This is due to the fact that the particles in a laser beam active area became hotter where the effect of recoil pressure becomes dominant causing the melt to flow up the sides of the kerf. As the kerf becomes deeper, the material is restricted by the kerf walls and the particles leave the kerf at a constant rate. Vaporisation was observed inside the kerf throughout the time. Overall, due to the vaporisation-induced recoil pressure and the effect of the assist gas, the melt and vapour simultaneously pushed
upward of the cavity whilst the melt front at the bottom propagated through the material thickness. The particles were started to be ejected downward when the full depth penetration occurred (both upward and downward material ejection occurred, which the material was solely ejected upward during the partial penetration depth) [215].

Furthermore, the present model was compared against the work done by Ng et al. [194]. The present model predicts a maximum melt ejection velocity up to 6.42 ms\(^{-1}\) with an average 5.1 ms\(^{-1}\) (Figures 8.11a and 8.11b) which is in acceptable agreement with data obtained by Ng et al. [194], of 8.5 ms\(^{-1}\). The deviation between the predicted results and the published data by Ng et al. [194] was due to the different operating conditions where 5 kW peak power of pulsed Nd:YAG laser has been utilised in their work.

![Figure 8.11: (a) Average particle velocity in a function material penetration depth obtained by the model for the full penetration depth](image)

8.5.4 Formation of recast and spatter

Recast is formed due to the molten material which is not blown away from the cut zone, and will re-solidify on the cut surface and form a recast layer. In a severe
condition, the recast may extend the heat to the parent materials and deteriorate the cut quality [216]. Work by Low et al. [217] demonstrated that the spatter formation characteristics were influenced by assist gas type, laser pulses and workpiece material composition and peak power. Low et al. [218] suggests that short pulse durations, low peak powers and higher repetition rate could lead to a small area of spatter.

Figure 8.12 shows the molten materials have accumulated on the side of the cut edge and re-solidify subsequently form a recast and spatter obtained experimentally. Figure 8.13 shows the formation and recast predicted by SPH model. Note, in Figure 8.13, the particles were coloured by temperature. It is important to mention that the experimental result for spatter and recast presented in Figure 8.12 was obtained with a moving beam while the predicted result by SPH model was just obtained with a static beam. This result is still far too coarse to be compared against the experimental data; however, this result is presented here in to show the potential capability of SPH in handling such problem in laser processing.

Figure 8.12: Example of recast and spatter formation on the cut surface obtained experimentally.
Chapter 8: Understanding the behaviour of pulsed laser dry and wet micromachining processes by multi-phase SPH

8.5.5 Backwall damage and effect of water flow in thin tube cutting

Wet cutting is preferable in stent processing to overcome processing issues such as back wall damage due to the melt ejection adheres to the opposite wall of the tube. This would cause rough surface finishes to the stent structures with undesirable consequences for operational use. The use of water flow inside the tube to assist the cutting process has been found to significantly improve back wall finish due to the water stream protect the back wall as well as assist in waste material evacuation.

Experimentally, water of thickness 2.87 mm was flowing inside the tube diameter, however in the modelling, 100 µm of water thickness has been utilised to minimise the simulation time. Herein wet cutting was modelled to show how the hydrodynamic effect of water in preventing the back wall damage to demonstrate how the water carries away the ejected particle and also cools down the particles. Figure 8.14 displays the snapshot of the melt ejection time history during the wet cutting (velocity of water, \( V_w = 0.241 \text{ ms}^{-1} \)). This demonstrates how the water flow carries away the ejected particles (shown in red in the blue water region). Soon after exiting the kerf, the ejected
particles fall in the water and were carried away from the ejected region which was sufficient for material evacuation.

No experimental data exists to validate the movement of the molten particle inside the water. However, the spatial distribution of the ejected solid material was plotted as shown in Figure 8.15. Additionally, the figure is also aiming to capture the behaviour of the molten particles after being ejected from the kerf in dry and wet cutting condition. In dry cutting, it was observed that the particles were ejected towards the backwall resulting in hot particles adhering and transferring heat to the back wall. During wet cutting, soon after exiting the kerf, the molten materials were entering the water and are carried away by the water stream from the region where it was being ejected. The continuous water flow kept removing the molten particles whilst protecting the back wall from those hot particles. It also can be observed that the hot molten particles start to cool down when in the water.

![Figure 8.14](image)

Figure 8.14: Snapshots of melt ejection time history during wet cutting (water velocity, $V_w = 0.241$ ms$^{-1}$), (a) 0.15 ms (b) 0.175 ms (c) 0.2 ms and (d) 0.25 ms. Note: the particles were coloured by density.
Figure 8.15: Ejected particles behaviour after ejected from the kerf in dry cutting and wet cutting, (a) at $t = 0.15$ ms (soon after ejected from the kerf) and (b) at $t = 0.25$ ms. Note: the particles were coloured by temperature. Cross section of the model is shown in z-y view for a clear capture of the molten materials.

The back wall for both machining conditions obtained by SPH model was then compared against the experimental data. In the SPH model, it can be observed that the ejected particles were scattered and adhered to the opposite wall as shown Figure 8.16a (top left) for the dry machining. In the wet machining, a desirable clean and spatter free back wall was obtained as shown in Figure 8.16b (top right) where the continuous water stream carries away the hot particles after they were being ejected out of the bottom of the kerf. The modelling results implied that water aids material evacuation and cooled the ejected particles thus providing good back wall quality. This also shows the significance of water flow for wet cutting which was in agreement with the empirical qualitative analysis as shown in Figure 8.16 (bottom).
Figure 8.16: Spatial distribution of the ejected particles deposited to the back wall in SPH simulation (top images) and SEM images zoom of backwall obtained experimentally (bottom images), (a) dry cutting and (b) wet cutting.

8.6 Conclusions

An SPH model has been described to understand thermodynamic and hydrodynamic phenomena in dry and wet pulsed fibre laser micro-cutting and drilling of stainless steel. Realistic results have been presented where the phase changes in laser micro-machining are being considered for the first time using SPH method. Being significantly different from traditional mesh-based methods, this methodology avoids the necessity of using a computational mesh. The model developed was based on the material removal mechanism of melt ejection and vaporisation for laser beam intensity within a millisecond regime. The developed model for penetration depth gives good agreement with the experimental data and the melt ejection velocity gives acceptable agreement with the published data. Formation of recast layer also was shown in this work. Incorporation of water in the model simulates the wet micro-machining behaviour where the role of water in carrying away the debris and resulting clean backwall damage was successfully demonstrated. Even though the developed model is
limited to a static beam, this contribution could lead to future research in laser processing and significantly expand the capability of SPH.
Chapter 9

Conclusions and Future Work

9.1 Conclusions

In this PhD project, an investigation was carried out to understand the feasibility and process characteristics of three novel laser stent cutting processes: water flow assisted fibre laser cutting, picosecond laser cutting and water assisted femtosecond laser cutting. A numerical model has been developed for understanding the laser and material interaction in laser cutting process by using SPH method. Individual conclusions were drawn at the end of each chapter for both the experimental and modelling work. For this final chapter key findings and technological and scientific contributions made by this research work are presented as follows:
9.1.1 Experimental work

9.1.1.1 Comparison of dry and wet fibre laser profile cutting of thin 316L stainless steel tubes for medical device applications

- The laser cutting of stainless steel 316L with fibre laser cutting within the millisecond pulse duration was demonstrated. It was found that at this beam/material interaction time scale, the primary material removal mechanism is melting with molten metal ejected by an assisting gas jet. Hence, significant amount of dross was found attached to the cutting edge, along with backwall damage due to particle deposition in the backwall of the tube, and poor surface quality.

- With the introduction of water through the tube during fibre laser cutting (known as wet cutting), significant improvement was achieved. Compared to the dry cut, wet cutting resulted in less dross and no backwall damage. On the other hand, lower surface roughness, smaller HAZ and also narrower kerf widths were observed with water assisted fibre laser cutting.

- Water assistance during the process had the role of carrying away the ejected particles before they permanently adhered to the backwall whilst cooling the workpiece and the ejected particles. It was found that water does not play a significant role in attenuating the laser beam when cutting small diameter tubes. In this work, it was determined that ~ 94% of the beam intensity reached the backwall in wet cutting. It was found that for $\lambda = 1080 \pm 5$ nm, 100% beam absorption can be obtained at around 4.2 cm water depth.
9.1.1.2 Picosecond laser micromachining of nitinol and platinum iridium alloy for coronary stent applications

- Picosecond laser cutting was demonstrated, for the first time, in cutting nitinol and platinum iridium alloy. It was found that a picosecond laser, operated at UV range ($\lambda = 343$ nm) and 6 ps pulse duration, offered better cutting quality (i.e. dross free cut, clean back wall, minimum HAZ and improved surface finish).
- Further studies in this work found that the picosecond laser has great potential for laser micro-machining of other stent materials including cobalt chromium, silver, titanium and polymer.
- Picosecond laser cutting delivered an acceptable cutting quality without using water assistance during the cutting process.

9.1.1.3 Underwater femtosecond laser micromachining of thin nitinol tubes for medical coronary stent manufacture

- Femtosecond laser cutting of nickel titanium tube was also demonstrated. From a general observation, it was concluded that cutting with a femtosecond laser resulted in the accumulation of debris and also the generation of a prominent recast layer.
- Underwater femtosecond laser cutting was performed, for the first time, to improve the cutting quality for stent cutting. It was observed that underwater cutting produced debris-free cuts with no recast, no HAZ and high surface quality.
- The loose debris can be removed with ultrasonic cleaning whereas the recast attached permanently to the material required subsequent processing in order to be removed.
- The use of compressed air (pressure = 2 bar) along the sides of the cutting direction did not help minimise debris and recast formation.
Both dry and underwater femtosecond laser cutting gave high surface quality at the optimum parameters. In the dry cutting, minimum surface roughness was obtained at low fluence (4 J/cm$^2$) and high speed (1.5 mm/s). For underwater cutting the optimum parameters changed to high fluence (13 J/cm$^2$) and low speed (0.1 mm/s) which also showed minimum surface roughness in the cut material.

9.1.2 SPH modelling

- SPH modelling was performed by using the open-source code (SPHysics) which successfully demonstrated the capability of this method towards the application in laser cutting process which involves melt splashing leaving the kerf which is difficult to model by conventional methods.
- It was found that the developed model for penetration depth is in good agreement with the experimental data and the modelled melt ejection velocity is also in accordance with published data.
- Formation of recast and spatter was also included in the result to show the capability of SPH in handling such problem in laser processing.
- Incorporation of a water environment in the model simulates the wet cutting behaviour where the role of water in cooling and carrying away the debris resulting in clean back wall damage was successfully demonstrated.
9.2 Recommendations for future work

The recommendations for future work that will aid to improve the process and enhance understanding of laser stent cutting are summarised as follows:

9.2.1 Experimental work

- The study of fibre laser cutting reported in this thesis was limited to stainless steel 316L samples, assisted by N\textsubscript{2} at fixed pressure. It is recommended to include a broader range of biocompatible materials with different type of assist gases at different pressures for further analysis of quality improvement. The results can be compared with the ones obtained with wet cutting to identify the optimum processing window in fibre laser cutting of stents.

- It was found that limited research available regarding the picosecond laser cutting particularly for stent applications. Research with picosecond laser cutting should be further investigated with different wavelengths and operating conditions to fill in the research gap in this area, particularly with a higher average powered ps laser.

- In the current study, a femtosecond laser was operated at fixed pulse duration (100 fs). The use of shorter pulses (few femtosecond) could achieve melting-free ablation; hence no recast layer formation is expected for a such range.

- Corrugated surface was observed in underwater cutting with femtosecond laser at high fluence and high cutting speed. It is suggested to use flowing water during the cutting process to reduce the localised bubble effect at the ablated surface to minimise this effect.

- Cleaner surfaces can also be expected from machining in vacuum conditions.
9.2.2 SPH modelling

- The three-dimensional SPH modelling presented herein is considered for the first time for a laser micro-machining and drilling process involving penetration depth, phase changes, melt ejection velocity, spatter and recast, and water assistance. Thus, a series of assumptions were followed for model simplification purposes. Assumptions in the model can be reduced by incorporating more mechanisms involved during the physical laser cutting process.

- In the current model, only static beam was utilised throughout the simulation. The model can be further developed to enhance the understanding of the physical process in laser cutting. It is recommended to follow the model approach in this work which can be further enhanced with the application of a moving beam source.

- Further model can be developed with a wide range of laser pulse durations, repetition rate, laser power and different assist gas.

- To increase the model accuracy, more empirical data should be obtained on a wider range of materials and operating parameters, i.e. laser power, pulse duration, gas pressure.

- For water assisted cutting, the current model only considered the existence of water underneath the workpiece. Further modelling can be developed for the submerged condition.

- It is suggested to produce a model with higher resolution for an enhanced and more accurate prediction. Numerical convergence study is also needed for model accuracy.

- The work could be extended to sensitivity analysis of the SPH model. This in particular could investigate the significance and rationale of the artificial viscosity. The broader aim would be to establish physics and process mechanism and understanding of the conditions that are required to create a stable model.
References


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