Medical Engineering and Physics 000 (2017) 1–11

Contents lists available at ScienceDirect

Medical Engineering and Physics

journal homepage: www.elsevier.com/locate/medengphy

Integrated experimental and computational approach to laser machining of structural bone

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ARTICLE INFO

Article history:
Received 18 July 2017
Revised 26 October 2017
Accepted 22 November 2017
Available online xxx

Keywords:
Laser
Machining
Structural bone
Orthopaedic

ABSTRACT

This study describes the fundamentals of laser–bone interaction during bone machining through an integrated experimental-computational approach. Two groups of laser machining parameters identified the effects of process thermodynamics and kinetics on machining attributes at micro to macro. A continuous wave Yb-fiber Nd:YAG laser (wavelength 1070 nm) with fluences in the range of 3.18 J/mm^2–8.48 J/mm^2 in combination of laser power (300 W–700 W) and machining speed (110 mm/s–250 mm/s) were considered for machining trials. The machining attributes were evaluated through scanning electron microscopy observations and compared with finite element based multiphysics-multicomponent computational model predicted values. For both groups of laser machining parameters, experimentally evaluated and computationally predicted depths and widths increased with increased laser energy input and computationally predicted widths remained higher than experimentally measured widths whereas computationally predicted depths were slightly higher than experimentally measured depths and reversed this trend for the laser fluence >6 J/mm^2. While in both groups, the machining rate increased with increased laser fluence, experimentally derived machining rate remained lower than the computationally predicted values for the laser fluences lower than ~4.75 J/mm^2 for one group and ~5.8 J/mm^2 for other group and reversed in this trend thereafter. The integrated experimental-computational approach identified the physical processes affecting machining attributes.

Abbreviations

CW:	Continuous wave
R_E:	Experimental ratio of depth to width of machined cavity
R_C:	Computationally predicted ratio of depth to width of machined cavity
k:	Thermal conductivity
C_p:	Specific heat
ρ:	Density
T:	Temperature
t:	Time
R:	Beam residence time
d:	Laser beam diameter
V:	Laser beam scanning speed
ε:	Emissivity
P_g:	Three-dimensional Gaussian laser beam distribution
σ:	Stefan–Boltzman constant
T_0:	Ambient temperature
M_r:	Recoil pressure
L_v:	Latent heat of vaporization
M_v:	Mass of vapor molecule
T_v:	Voporization temperature
T_s:	Instantaneous surface temperature
B:	Fraction of Particular phase (solid, liquid or vapor)
T_x:	Transition temperature between two distinct phases (solid, liquid or vapor)
α:	Thermal expansion coefficient
g:	Gravitational acceleration

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https://doi.org/10.1016/j.medengphy.2017.11.010
1350-4533/

h:	Heat transfer coefficient
Yb:	Ytterbium
Nd-YAG:	Neodymium-doped yttrium aluminium garnet
Ho:YAG:	Holmium-doped yttrium aluminium garnet
Er:YAG:	Erbium-doped yttrium aluminium garnet
ESEM:	Environmental scanning electron microscope
EDS:	Energy dispersive spectroscopy
F:	Laser fluence
P_0:	Average laser power
A:	Cross-sectional area of laser beam on target surface
t_0:	Beam residence time

1. Introduction

An osteotomy is one of the bone shaping techniques used during orthopaedic surgeries. Orthopaedic surgical techniques have evolved through adaptation and integration of sensors and CAD based generation of patient specific instrumentation and bone cutting parameters. However, surgeries are still largely done by surgeons using conventional mechanical tools such as saw, osteotome, and burr [1]. Hence, conventional orthopaedic surgery depends on human and tool attributes resulting in poor reproducibility and thermally driven necrosis due to friction/abrasion between cutting tool and bone for extended period [2–6]. This provides an opportunity for development of operating tools and techniques to address the adverse effects of orthopaedic surgery, such as (1) damage of tissues within and surrounding operated region, (2) low dimensional precision in operated bone, (3) slow surgical process, (4) post-surgery tissue trauma, and (6) cost of revision surgery.

In general, saw blades are harsher than burrs [1, 2] with temperatures of bone rising above 100 °C due to the large contact area of bone with the saw teeth and repeated cuts required to shape the bone. Although the burr cutting results in moderate temperatures (50–60 °C) [6], it is limited to shallow cuts. Many remedies have been explored to address temperature rise, and associated necrosis, including change in tool design [7, 8], and employing saline cooling [9, 10]. Internally cooled cutting tools provided effective heat control than an external spray/mist cooling [8, 10]. Also, due to physical contact between the mechanical cutting tools and bone, sterilization of tools is necessary.

Due to the composite nature of bone, the direction of mechanical loading critically determines the response of bone to cutting. Factors such as porosity, mineralization, orientation/diameter/spacing of collagen fibers, and histological structure of bone play a deterministic role in mechanical response [11, 12]. Thus anisotropy and heterogeneity of bone makes the exact response of the cutting process difficult to predict leads to generation of site specific stress concentration, cracks/microfissures, fracture [13], and poor quality of cut bone surface. These factors also have a significant bearing on post-operation bone ingrowth and healing characteristics [14, 15]. Therefore, it is extremely difficult to select cutting parameters (speed, force, feed rate) and saw parameters (tooth spacing/pitch, tooth size, tooth form) for desired outcome. Subsequently, the quality of the outcome mostly remains an art related to skill and experience of the surgeon.

Based on the above discussion, it is evident that an osteotomy needs to aim at the elimination of complex operating tools, variables of human factor, and physical contact between the cutting tool and bone. In view of this, the present effort employed a novel and non-conventional laser based non-contact technique for bone machining. The technique employed short duration high intensity laser energy for machining. The laser machining affords inherent advantages such as tight control over processing parameters via computer based automation for precision, repeatability, confined heating for minimal thermal damage in surrounding volume, and speedy operation [4]. Furthermore, being a non-contact method, laser machining eliminates mechanical loading and cracking associated with the residual stresses. These primary advantages are expected to lead to the benefits such as rapid patient recovery and reduced operating cost.

Lasers have been previously explored for ablation of bone and other hard tissue [16–42]. Most of these studies were confined to evaluation of correlations between laser parameters and resultant morphology of ablated surface (depth of ablation), ablation rate (machining rate), and thermal effects (necrosis and microfissures). Other few studies were focused on machining of hard tissues such as monolithic dental enamel in athermal (cold ablation) manner (without collateral damage to the tissue). However, cold ablations produced very shallow depths ($\leq$ 1 μm) and/or low machining rates (<1 mm^3/s) due the need to use of ultrashort femtosecond (fs) or picosecond (ps) pulse lasers [16–19]. Although remaining studies [20–41] employed continuous wave (CW) and pulsed CO_2, Nd:YAG, Ho:YAG, and Er:YAG lasers for machining of bones, they were confined to only shallow drilling and cutting (<2 mm depth) operations. Observations of these studies, repeatedly indicated generation of very low machining rates (1 mm^3/s) and substantial thermal necrosis of bone around the cut region. On the contrary, in orthopaedic surgeries for hard tissue implant replacement (knee and hip), higher machining rates ($\geq$50 mm^3/s) with minimal or no collateral thermal damage to the structural bone is desired. Hence, the current work presents the preliminary efforts on machining of structural bone using Yb-fiber coupled Nd:YAG laser (1.07 μm wavelength). The efforts were primarily focused on fundamental understanding of laser interaction with bone. Such a basic understanding was attempted with an integrated experimental and computational approach for optimization of laser process parameters to achieve the most effective removal/ablation (shaping/cutting) of the bone with least collateral thermal damage.

2. Materials and methods

2.1. Material

Fresh bovine cadaver femur specimens were collected from a slaughterhouse. The mid-shaft portions were isolated and sectioned into 100 × 25 × 20 mm blocks of cortical bone using a commercial band saw. The specimens were ultrasonically cleaned in normal saline for one hour followed by a cleaning cycle in distilled water-formaldehyde solutions in volumetric proportions of 75−25%, 50−50%, 25−75%, 15−85%, 10−90%, and 5−95% with each cycle lasting 12 h. The process was concluded with a 12-h immersion cleaning in 100% formaldehyde. This removed all soft tissue and cartilage externally attached to the bone. The sample surfaces to be laser machined were lightly ground on 800 and 1200 grit papers for uniformly flat and smooth (~3 μm average roughness) surface and cleaned with distilled water to remove lose debris before laser machining. The cleaned specimens were air-blow dried for 10 min, sealed in plastic containers, and refrigerated until they were subjected to laser machining within 24 h.

2.2. Laser machining

A CW Yb-fiber coupled Nd:YAG laser (1.07 μm wavelength) was employed to conduct laser bone machining trials. To understand the fundamentals of laser–bone interaction a single laser track was produced with each set of machining parameters. The laser beam diameter on sample surface was 0.6 mm. Laser power, and scanning speed in current efforts were 300–700 W and 100–250 mm/s, respectively. The resultant laser fluence and beam resident time ranged over 3.18–8.48 J/mm^2 and 2.4–6.0 ms, respectively (Table1). The machining trials were conducted in argon cover gas flown at 3 l/min to avoid oxygen contamination of the specimen surface.