Effect of Nd-YAG laser treatment on corrosion behavior of AISI316L stainless steel in artificial saliva solution

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Abstract. This work aims to study the effect of Nd:YAG laser treatment on surface modification and corrosion behavior of AISI 316L stainless steel which is the most widely used as implant bio material in the artificial saliva solution at 37 ±1ºC. The samples were irradiated with Nd:YAG laser with various pulse repetition rates (or frequencies) of 1, 3 and 6 Hz and with two wavelengths of 1064 nm and 532 nm at fixed energy of 1000 mj. Microstructure examinations by optical and SEM were performed before and after laser treatment. Corrosion tests were carried out by using potentiostat instrument and the corrosion rate was measured by Tafel extrapolation method, also cyclic polarization was done to study the active-passive behavior of stainless steel 316L in saliva solution. It has been found that the treated samples at different laser conditions have hardness higher than that of untreated sample. The results showed that the hardness increased slightly with decreasing the pulse repetition rate or frequency from 6 to 1 Hz at 1064 nm, also a surface modification and grain refinement were noticed. Corrosion results indicated that the treated samples with 1 Hz with 532 nm and 1064 nm showed higher corrosion resistance than untreated base alloy and other treated samples.

1. Introduction

The increasing request of laser material processing can be attributed to several distinctive properties of laser namely, high production non-contact processing, reduced of finishing operation and processing, low cost, improved quality of product, greater material using and minimum heat affected zone [1,2]. In general, application of laser material processing can be categorized into two major classes, (a) applications requiring limited energy or power and causing no major change of phase and (b) applications requiring significant amount of energy to induce the phase transformations. The first category includes semiconductor annealing and etching, polymer curing, scribing/marking of integrated circuit substrates, etc. The second category includes cutting, welding, melting, heat treatments, etc. [3]. Failure of engineering materials can be classified into several types: corrosion, oxidation, fatigue and wear/abrasion are most likely to initiate from the surface because of two main causes: 1- free surface is more prone to environmental degradation, and 2- intensity of externally applied load is often highest at the surface. There are many engineering solutions to minimize or eliminate such surface initiated failure, which lies in modification of surface composition and/or microstructure of the near surface region of a part or piece without affecting the bulk [4,5]. These engineering solutions are, the more commonly practiced conventional surface engineering techniques such as galvanizing, diffusion coating, carburizing, nitriding and flame/induction hardening which possess several limitations like high time/energy/material consumption, poor precision and flexibility.
Stainless steels, Ti alloys and Co-alloys, are widely used in biological implant materials for skeletal load bearing applications. Metals and their alloys have played an important and the main role in structural biomaterials in reconstructive surgery, especially in orthopedics. The corrosion was occurred on the surface of these metal and its alloy, so laser was used to modify the mechanical properties such as hardness. Most of laser surface treatments of stainless steel have been focused on martensitic (420), ferritic (430) and austenitic types (304 and 310) to improve pitting corrosion resistance. A 316L austenitic stainless steel is the most commonly used implant materials as it is cost effective [6]. Stainless steel 316L has high corrosion resistance which is attributed to the formation of adherent and coherent oxide thin film of chromium oxide (Cr2O3) on the surface and serves as an obstacle to corrosion reactions. But this chromium oxide passive film formed on the surface of the implant metal is unstable and prone to localized corrosion in longtime implant applications due to cruel biological effect because the human body is a complex and aggressive electrolytic environment for biomaterials applications [7,8]. There are many researches and studies made in the field of enhancing the surface characteristics such as hardness, wear and corrosion resistances of dental alloys and alloys by laser surface treatment because of their importance in life.

Khairfallah et al.[9] (2011) investigated the surface modification of stainless steel 316L SS by laser melting with using 2 and 4 KW laser power type (CO2) with different ranges of scanning speeds (300-1500) mm/min. They added carbon in same material by using alloying laser surface which includes pre coating the sample surfaces with graphite powder and then followed by laser melting. The results showed that the laser treatments are to enhance corrosion resistance and hardness of stainless steel surface. This is due to the fine and homogenous microstructure in the melted zones. Freire et al. [10] (2012) studied the electrochemical corrosion behavior of AISI 316L and AISI 304 stainless steels by using potentiodynamic mode by cyclic voltammetry. Chemical compositions of passive films which formed by cyclic voltammetry and the behaviour of electrochemical were compared with films that grown under ordinary conditions. Also the effect of laser surface modification on corrosion resistance of dental alloys (Co-Cr-Mo) and (Ni-Cr-Mo) were studied by Ataiwi et al. [11]. They used different laser energies in order to increase resistance of corrosion for the alloys which used dental applications in artificial saliva with having alcohol beverages that detected the corrosion behavior of dental alloys.

The electrochemical corrosion for laser melted layer to stainless steel AISI 321 in artificial saliva was performed by Dikova et al. [12] Xiaoyan Zhang et al. [13](2017) studied the effects of laser heat input on the microstructure, hardness and corrosion resistance of austenitic stainless steel AISI 316L treated by laser heat treatment. They noticed that the laser surface treatment is refined grain and increased the hardness of 316L orthodontic bracket at certain heat input. Recently Rasheed et al. [14] (2016), Abbass and Abdulkareem [15] (2018) studied the effect of Nd-YAG laser energy on the microstructure and corrosion behavior of AISI 316L stainless steel which is the most widely used as surgical implant material in the artificial saliva solution. The results showed that corrosion resistance was improved after laser surface treatment due to surface modification and microstructure refinement.

The objective of this work is to use a novel technique as an attempting to improve the corrosion resistance of stainless steel AISI 316L used as dental alloy in the artificial saliva solution by using Nd-YAG laser surface treatment with various pulse repetition rates of (1, 3 and 6Hz) and two wavelengths of laser beam (1064nm and 532nm) with fixed pulse energy (1000mj).

2. Experimental work

2.1. Material

AISI 316L austenitic stainless steel is used in this work due to its wide applications in biomedical field and it is used as orthopedic and dental implants. This steel was as rolled sheet with 1.0mm thickness. The chemical composition of AISI 316L was evaluated using spectrometer analysis instrument available in the Central Organization for Standardization and Quality Control –Iraq as shown in Table 1. Several samples with surface area of 1.5 cm x 1.5 cm were prepared using a wire cutting machine. After that the surfaces were ground with different grades :320, 500 and 1000 of SiC emery paper. Then polishing process was performed using a special cloth with diamond paste of 0.3 µm and oil to make the surface flat and polish.
Table 1. The chemical composition of stainless steel AISI 316L.

| Element (wt%) | C | Cr  | Ni  | Mo | Mn | S  | P  | Fe  |
|--------------|---|-----|-----|----|----|----|----|-----|
| Standard     | 0.03 | 16.18 | 10.14 | 2.3 | 2.0 | 0.03 | 0.045 | bal |
| Analytical   | 0.029 | 16.17 | 9.82 | 2.0 | 1.34 | 0.04 | 0.03 | bal |

2.2. Laser Surface Modification (LSM)
Nd:YAG laser (Q switch laser) instrument was used in this study as shown in Figure 1. Table 2 indicates the specifications of Nd:YAG laser which used in this study. Area of 1 cm² from sample was irradiated by a laser beam at distance of 8 cm away from the lens with spot size of 2mm. It was used with different pulse repetition rates of (1, 3 & 6 Hz) and two wavelengths of (1064 and 532 nm) with fixed pulse energy of 1000 mj (power intensity = 0.318 J/mm²) and pulse duration of 6 ns were used for all samples to modify stainless steel surface and increase the hardness and corrosion resistance for used in dental implant; then the corrosion test was carried out with sample put in artificial saliva solution as shown in Table 3.

![Nd:YAG laser instrument](image)

Figure 1. Shows Nd:YAG laser instrument used in the present work.

Table 2. Specification of Nd:YAG laser instrument used in this study.

| Laser type       | Nd:YAG Q switch laser       |
|------------------|-----------------------------|
| Wavelength       | Double wavelength (1064,532) nm |
| Controller       | 7.4 color touch LCD display |
| Pulse energy     | 1600 mj                     |
| Power            | 1000 w                      |
| Width of pulses  | 6 ns                        |
| Frequency        | 1-6 Hz                      |
| Spot diameter    | 1-7 mm adjustable           |

2.3. Microstructure and Microhardness Test
Microstructure examination was carried out for laser treated (AISI316L) specimens. After sample’s preparation by grinding and polishing processes as mentioned above etching process has been done by immersing the samples in etchant solution consisted of (8 ml HCl+ 6 ml HNO₃+ 2 ml glycerol + 4 ml H₂O). Then the sample was washed by water and cleaned by acetone then dried by hot air. The
microstructure examination has been implemented by using Metallurgical light microscope (Type MBL3300) (model KRUSS Optronic- Germany) connected to computer.

Scanning electron microscope (SEM) type (VEGA3LM TESCAN COMPANY) provided with EDS (type Oxford) was used to examine the microstructure and topography of sample surface. XRD analysis was performed using X-Ray diffractometer type (Shimadzu- XRD-6000) Japan using Cu-tube with wavelength 1.540 Å, current 30 mA, voltage 40KV and scan rate was 5° / min to identify the main phases in the untreated AISI 316L sample.

The Vickers micro hardness test of laser treated samples was carried out by using Digital MicroVicker Hardness Tester type (TH714) - China. A load of 500 Kg has been applied for holding time of 30 Sec. The 3-5 reading of hardness values were done and average was taken.

2.4. Preparation of corrosion solution

Artificial saliva solution was prepared in laboratory and its chemical composition is shown in Table 3. The pH of solution was measured by pH meter and it was 5.3.

Table 3. Chemical composition of the artificial saliva solution.

| Material | NaCl | KCl | CaCl2.2H2O | NaH2PO4 | Na2S | KSCN | Urea |
|----------|------|-----|------------|---------|------|------|------|
| Concentration g/ L | 0.4  | 0.4 | 0.78       | 0.69    | 0.005| 0.3  | 1.0  |

2.5. Electrochemical corrosion Cell

A sample of an exposed area (1 x 1 cm2) is mounted with holder and immersed in the electrochemical corrosion cell which consists of three electrodes. The standard hydrogen electrode (SHE) was used as reference electrode and the auxiliary electrode was made of platinum. The holder of sample (working electrode) together with the reference and auxiliary electrode were putted in the respective positions in the electrochemical cell used for this work as shown in Figure 2. All data were recorded after one hour until the potential value reaches to a steady state value which represents the open circuit potential (OCP). A current reading of cell is taken for a short, slow drag of the potential. The drag was taken from (–200 to +200) mV relative to (OCP). A data for linear fitting of the standard model is given as an estimation of the polarization resistance, that calculated the corrosion current density (Icorr) and rate of corrosion (mpy). The investigation is achieved during a WENKING MLab multi channels and SCI-Mlab corrosion measuring system from Bank Electronics- Intelligent controls GmbH, Germany 2007. From corrosion tests for all the samples in the saliva solution at 37 ±1 ºC with pH of 5.3, the corrosion parameters, (corrosion potential (Ecorr) and corrosion current (Icorr)) were determined by Tafel extrapolation method. Corrosion rate is calculated from the equation 1[16]:

\[ C.R \text{ (mpy)} = 0.13 \times I_{corr} \times E.W / \rho \] (1)

where; I corr ; is corrosion current density (μA.cm-2), E.W ; is equivalent weight (gm), and \( \rho \); is the density (gm/cm3). mpy meaning mils per year (0.001 in ) as a penetration range.

Polarization resistance (Rp) can be determined from Tafel slopes according to Stern-Geary equation 2 [16].

\[ Rp = (bc.ba) / (2.303 \times I_{corr} \times (bc+ ba)) \] (2)

Where: bc , ba; are cathodic and anodic slopes (mV.dec-1) , Rp; polarization resistance (Ω .cm2) and I corr ; is corrosion current density (μA.cm-2).
3. Results and Discussion

3.1. Microstructure
Optical micrograph or microstructure of stainless steel AISI 316L as received is shown in Figure 3a which consists of $\gamma$-austenite phase containing twins. While XRD results confirmed the presence of $\gamma$-austenite phase in different Bragg angles (2Theta angles) in stainless steel as shown in Figure 3b.

Figures 4, 5, 6 and 7 show the microstructures of laser treated sample at 1000mj with different pulse repetition rates and two wavelengths of 1064 and 532nm. It was observed three zones in laser treated sample varied in grain size and a fine dendritic structure homogeneously distributed in the melted zone as shown in Figures 4 and 5.
laser surface treatment of AISI316L stainless steel at different pulse repetition rates and wavelengths at fixed pulse energy of 1000mj results in melting the surface layers of metal in laser zone and produce a fusion zone resembles as small cast and gave a microstructure consisting of columnar or longitudinal grains grown from the interface toward the surface. Also it was seen fine dendrites within these grains as shown in (Figures 6 and 7). This is due to a very high cooling rate during solidification.

SEM-EDS images confirmed these results for sample treated with fixed frequency of 3Hz and 1064nm and 532nm as shown in Figures 8 and 9 respectively. When the wavelength ($\lambda$) reduces, the energy increases according to equation ($E = hc/\lambda$) and then heat input increases that led to more surface melting of metal in localized small region results in rapid solidification forming dendritic structure.
Figure 8. SEM micrographs images for the sample AISI 316L after laser surface treatment at 1064nm and 3Hz with fixed energy 1000 mJ (a) 50 µm, (b) 20 µm, (c) EDS elemental chemical analysis of the laser zone in the stainless steel 316L.

Figure 9. SEM micrographs images for the sample AISI 316L after laser surface treatment at 532 nm and 3Hz with fixed energy 1000 mJ (a) 20 µm, (b) 20 µm, (c) EDS elemental chemical analysis of the laser zone in the stainless steel 316L.
3.2. Microhardness

Table 4 indicates the results of microhardness values for untreated base alloy (316L) and treated samples after irradiate with Nd-YAG laser beam at fixed pulse energy (1000mj) and at three pulse repetition rates of (1, 3 and 6 ) Hz and two wavelengths of (1064 and 532) nm. It has been found that the treated samples at different conditions have hardness values higher than that of untreated specimen. This is due to the interaction between laser pulses and stainless steel that lead to high heat input and thermal gradient, hence to rapid solidification and fast cooling rate. Also surface modification and fine grains of γ-austenite phase in laser zone were observed for surface laser melting produced parts.

It was seen that the hardness increased slightly as the pulse repetition rate decreased from 6Hz to 1Hz at 1064nm. That means with increasing the pulse repetition rate the amount of heat increased and it makes the metal (stainless steel) retain part of heat in a short time periods and not sufficient to dissipate the heat. Also the temperature of metal does not reach room temperature that results to heat accumulated in the metal and led to low cooling rate. The microhardness in laser spot or in melted zone was higher than that of the heat affected zone and base metal.

The wavelength of laser beam also affects on the interaction of laser beam with material and the process of laser beam. It is determined the amount of absorbed laser energy falling in the metal. It is better to use short wave length as possible in most industrial and medical applications. In this study the samples treated with green light (532nm) are exhibited hardness values higher than the untreated base alloy because the decrease in wavelength of laser beam led to increase in energy and heat input become sufficient to melt metal in small laser zone or laser spot shows a refine grains and a high interdendrite percentage regions resembling a eutectic structure grains which increased the microhardness to higher values than untreated metal. These results are in agreement with results of researcher Khalfallah et al. [9].

| Sample Symbol | Laser treatment conditions | Hardness HV Kg/mm² | Improvement % |
|---------------|-----------------------------|--------------------|---------------|
| A             | Untreated base 316L         | 162                | -             |
| B             | 1000mj, 1Hz, 1064 nm        | 195.4              | 20.6          |
| C             | 1000mj, 3Hz, 1064 nm        | 194                | 19.7          |
| D             | 1000mj, 6Hz, 1064 nm        | 191                | 17.9          |
| E             | 1000mj, 1Hz, 532 nm         | 179.2              | 10.6          |
| F             | 1000mj, 3Hz, 532 nm         | 182.6              | 12.7          |

3.3. Corrosion behavior

Corrosion refers as the destructive and damage of a metal and the electrochemical process at the surface which is of great importance mainly when a metallic implant is placed in the aggressive electrolytic environment of the human body. The implants and dental alloys facade rigorous corrosion medium which includes blood and numerous constituents of the body fluid as in the case of saliva solution. The aqueous medium in the human body consists of various anions such as chloride, phosphate and bicarbonate ions, cations like Na+,K+,Ca2+,Mg2+ etc. [17].

3.3.1. Open circuit potential (OCP)

Figure 10 shows the OCP-time measurements for untreated 316L stainless steel and treated sample of dental alloy in artificial saliva at pH= 5.3 and 37 ± 1°C. It begins at potential -278 mV and then increased up to -248 mV because of forming Cr2O3 on the metal surface and stable at this value for 60min. While in case of treated sample at 1 Hz and 1064nm the OCP begins at -81mV and then dropped to -120 mV due to gradually breakdown of passive layer and stabilize to this value. The
sample treated at 1Hz and 532mV begins at -354 mV and then stabilize at -375 mV. This due to the stability of passive film in this alloy. In general the Eocp values were shift to the noble direction (less negative ) than untreated sample ( sample A ) except sample E and F . This is attributed to the formation of passive film that acts as a barrier for metal dissolution and reduces the corrosion rate as shown in Table 4. These results are confirmed by many researchers [18-20 ].

![Figure 10. OCP-time measurements for untreated and treated sample 316L at different laser conditions in artificial salvia solution at 37 ± 1º C.](image)

3.3.2. Cyclic polarization
Figures 11 (a,b, c, d and e) show the cyclic polarization curves for untreated and laser treated sample with different pulse repetition rates of ( 1 ,3 and 6 Hz) after corrosion test in artificial salvia solution at 37 ± 1º C. Figures 12(a,b,c,d and e) show the cyclic polarization curves for untreated and laser treated sample with two laser wavelengths ( 1064 and 532nm ) and two pulse repetition rates of ( 1Hz and 3Hz) after corrosion test in artificial salvia solution at 37 ± 1º C. It was seen that these anodic polarization curves consist of active- passive region and repassivation region until reach breakdown the passivity on alloy surface. These results are similar with researchers [14,15]. It can be seen from Figure 11 that treated sample of 316L have the same behavior in cyclic polarization curves as in Figure 12 which is due to the similar corrosion mechanism in both laser conditions at fixed energy of 1000 mj. The untreated base metal shows larger hysteresis loop than treated sample (Figure 11 a and d ) while the samples treated with 1064 nm and at 1Hz and 3Hz as shown in (Figure 11 b and c ) respectively show wide passivation region and repassivation during backward scan in anodic curve with no hysteresis loop and the pits disappear from the sample surface.
Figure 11. Cyclic polarization curves for untreated and laser treated sample with different frequencies in artificial salvia solution at 37 ± 1 °C.
Figure 12. Cyclic polarization curves for untreated and laser treated sample with different laser conditions in artificial salvia solution at 37 ± 1 °C.
Table 5. Corrosion parameters results after corrosion test in artificial salvia solution at 37 ± 1 ºC.

| Sample Symbol | Laser treatment conditions | OCP (mV) | E corr. (mV) | I corr. (µA/cm²) | ba (mV/Dec) | bc (mV/Dec) | Rp (Ω.cm²) Corr. rate (mpy) | Improvement % |
|---------------|-----------------------------|---------|-------------|-----------------|-----------|-----------|---------------------------|---------------|
| A             | Untreated base 316L         | -275    | -864.4      | 15.48           | 486.8     | -444.3     | 6.5157                    | 6.786         |
| B             | 1000mj, 1Hz 1064 nm         | -120    | -558.5      | 0.754           | 140.5     | -116.7     | 36.712                    | 0.330         | 95          |
| C             | 1000mj, 3Hz, 1064 nm        | -248    | -329.1      | 6.24            | 304.2     | -315       | 10.768                    | 2.735         | 59.7        |
| D             | 1000mj, 6Hz, 1064 nm        | -225    | -404.8      | 21.37           | 640       | -558.2     | 0.0203                    | 9.369         | Not         |
| E             | 1000mj, 1Hz 532 nm          | -375    | -385.8      | 0.826           | 123.3     | -143.5     | 34.862                    | 0.362         | 94.6        |
| F             | 1000mj, 3Hz, 532 nm         | -278    | -351.6      | 16.75           | 1157.6    | -1051.3    | 14.282                    | 7.343         | not         |

Table 5 displays the corrosion results for all samples after Tafel polarization test in artificial saliva solution at 37 ± 1 ºC. The results showed that the corrosion resistance of 316L alloy increases significantly after laser surface treatment with pulse repetition rate of 1Hz with 1064 nm and 532nm (sample B and E) respectively in which that laser radiation gives a smoother surface and low corrosion current densities (Icorr). The resistance polarization (Rp) can be determined from the Tafel slopes (ba and bc) according to Stern-Geary equation (2) [16], and the values of Rp are shown in Table 5. It was seen that these anodic polarization curves consist of active- passive region and repassivation region until reach breakdown the passivity on alloy surface. These results are similar with researchers [14,15]. It can be seen that the sample treated with laser of 1064nm and 532 nm at 1 Hz exhibited the highest corrosion resistance than other treated samples and the improvement in corrosion resistance (reduction in Icorr. ) were 95% and 94.6% respectively. This is due to formation of a protective layer of chromium oxide on the surface which reduces the corrosion rate and improved the pitting corrosion resistance.

3.3.3. Optical Micrographs after Corrosion

Figures 13(a,b,c) and d show the optical micrographs images for the laser treated sample at fixed energy of 1000mj after corrosion test in the artificial saliva solution. There are many large and deep pits in the base metal (Figure 13 a) while it was observed shallow pits and very small pits distributed in metal surface of laser treated sample with different frequencies and wavelength as shown in (Figures 13 b and c). While there are no pits in sample treated with 1 and 3Hz at 1064nm as shown in (Figure 10d) because of presence of the passive protective oxide film of Cr2O3 on the metal surface. Generally, microstructural examination after corrosion experiments in saliva solution showed less pitting sites in laser treated samples than untreated surfaces. Ram Kishor Gupta et al. [21] confirmed these results in their work. The results of bio corrosion study are important with respect to medical implants and their biocompatibility in the human body [21, 22].
Figure 13. Optical micrographs images for the untreated and laser treated sample at different laser conditions after corrosion test in the artificial saliva solution.

4. Conclusion

Laser surface melting treatment of AISI 316L austenitic stainless steel using Nd-YAG allows obtaining a homogeneous modified surface structure with a very fine dendritic structure or grains of γ-austenite phase in melted laser zone. The results showed that the hardness increased after laser surface treatment of AISI316L irradiated with fixed energy (1000 mj) at different pulse repetition rates of (1,3 and 6) Hz with (1064 and 532) nm to reaches to maximum value of heat input. It has been found that the sample treated with pulse repetition rate of 1Hz with 1064nm show higher hardness value than the base alloy and other treated samples and the improvement in hardness was 20.6%. The increase in pulse repetition rate led to a slight decrease in HV hardness from 195.4 to 191, but the hardness values are decreased by moving away from center of the laser spot or fusion zone.

The corrosion results indicated that the sample treated with laser of 1064nm and 532 nm at 1 Hz gave better corrosion resistances and the improvement in corrosion resistance (reduction in $I_{corr}$) were 95% and 94.6% respectively, which confirms the formation of a protective layer of chromium oxide on the surface which reduces the corrosion rate and improved the pitting corrosion resistance. Also these samples exhibit less negative potentials in corrosion potential ($E_{corr}$) values than the untreated sample and other treated samples. There is a shift in $E_{corr}$ to more noble direction. Cyclic polarization results showed in wider passive–repassive range in laser treated sample of 316L alloy at different conditions with smaller or no hysteresis loop compared with untreated sample.

References

[1] Steen W M 1991 Laser material processing NewYork: Springer Verlag.
[2] Duley W W 1986 Laser surface treatment of metals, NATO-ASI Series (E) No.: 115 (eds)CW Draper, P Mazzoldi (Boston: Martinus Nijhoff).
[3] Mordike B L 1993 Materials science and technology (eds) RW Cahn, P Haasen, E J Kramer (Weinheim: VCH) 15: 111.
Draper CW 1980 Laser and electron beam processing of materials, CW White, P S Peercy (NewYork: Academic Press).

Mazumdar J 1983 Lasers for materials processing, M Bass (New York: North Holland).

Sivakumar M, Mudali K U, Raj eswari S 1994 Investigation of failures in stainless steel orthopaedic implant devices, fatigue failure due to improper fixation of a compression bone plate. J. Mater. Sci. 13 142-145.

Ali Parsapour, Khorasani S N and Fathi M H 2012 Effect of surface treatment and metallic coating on corrosion behavior and biocompatibility of surgical 316L stainless steel implant. J. of Mater. Science & Technology 28 125-131.

Yoshimitsu Okazaki, and Emiko G 2005 Comparison of metal release from various metallic biomaterials in vitro. J. of Biomaterials 11–21.

Khalfallah I Y, Rahoma M N, Abboud J H, Benyounis K Y 2011 Microstructure and corrosion behavior of austenitic stainless steel treated with laser. Optics & Laser Technology 43 806-813.

Freire L, Catarino M A, Godinho M I, Ferreira M G S, Simões A M P, Montemor M F 2012 Electrochemical and analytical investigation of passive films formed on stainless steels in alkaline media. Cement & Concrete Composites 34 1075–1081.

Ataiwi A H, Amaee R A, Mohsan A A 2013 Effect of laser surface modification on the corrosion resistance of dental alloys in artificial saliva containing alcoholic. J. Laser. Baghdad University Part A 12 43-52.

Dikova T, Tsaneva D, Ilieva M, Panova N, Galunska B 2015 Investigation of the electrochemical corrosion of laser melted layers of stainless steel in artificial saliva. Advances in Materials and Processing Technologies 1 115-123.

Xiaoyan Zhang, Yu Song, Xinhong Wang 2017 Microstructure and Corrosion Properties of Orthodontic Brackets by Laser Treatment. Int. J. Electrochem. Sci. 12 32–39.

Rasheed N K, Hubeatir K A, Hmood A F 2016 Improvement of corrosion resistance of dental alloys in oral environment at different temperature by laser irradiation Australian Journal of Basic and Applied Science 10 162-170.

Abbass M K and Abdulkareem M H 2018 Effect of laser energy on the microstructure and corrosion behavior of AISI 316L stainless steel in artificial saliva solution. A Scientific Refereed Periodical Journal 1 27-46.

Korb L J, Rockwell International and David L Olson 1992 Corrosion Volume 13 of the 9th Edition, 1978, Metals Handbook, Fourth printing.

Geetha M, Durgalakshmi D, Asokamani R 2010 Biomedical Implants: Corrosion and its Prevention - A Review Recent Patents on Corrosion Science 2 40-54.

Haitham M A, Muna K A and Sami A A 2014 Study of corrosion resistance of Co-Cr-Mo surgical implants alloy in artificial saliva. Iraq Eng. &Tech. Journal 32A 10.

Kemin Z, Jianxin Z and Thierry G 2006 Improved pitting corrosion resistance of AISI 316L stainless steel treated by high current pulsed electron beam. Surface & Coatings Technology 201 1393-1400.

Mohit Sharma A V, Ramesh K, Nirdhi S, Nidhi A and Bobin S 2008 Electrochemical corrosion behavior of dental/implant alloys in artificial saliva. ASM International 17 695.

Gupta, Neha Prasad, Arun Kumar Rai, Biswal R, Sundar R, Bose A, Ganesh P, Ranganathan K, Bindra K S, Kaul R 2018 Ramkishtudy on laser shock peened 316L stainless steel in simulated body fluid and corrosion chloride medium. Lasers in Manufacturing and Materials Processing. Springer. 8 (3) 270–282.

Muna K, Abbass, Sami A, Ajeel and Haitham M. Wadullah 2018 Biocompatibility, bioactivity and corrosion resistance of stainless steel 316L nano-coated with TiO2 and Al2O3 by atomic layer deposition method, The Sixth Scientific Conference "Renewable Energy and its Applications. Journal of Physics. IOP Conf. Series 1032 (2018) 012017,1-15. doi :10.1088/1742-6596/1032/1/012017.