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Proceedings of the 23rd International Congress on Applications of Lasers and Electro-Optics 2004

JOINING CHALLENGES IN THE PACKAGING OF BIOMEMS

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Abstract

Micro-joining and hermetic sealing of dissimilar and biocompatible materials is a critical issue for a broad spectrum of products such as micro-electronical, micro-optical and biomedical products and devices. Novel implantable microsystems currently under development will include functions such as localized sensing of temperature and pressure, electrical stimulation of neural tissue and the delivery of drugs. These devices are designed to be long-term implants that are remotely powered and controlled. The development of new, biocompatible materials and manufacturing processes that ensure long-lasting functionality and reliability are critical challenges. Important factors in the assembly of such systems are the small size of the features, the heat sensitivity of integrated electronics and media, the precision alignment required to hold small tolerances, and the type of materials and material combinations to be hermetically sealed.

Laser micromachining has emerged as a compelling solution to address these manufacturing challenges. This paper will describe the latest achievements in microjoining of non-metallic materials. The focus is on glass, metal and polymers that have been joined using CO2, Nd:YAG and diode lasers. Results in joining similar and dissimilar materials in different joint configurations are presented, as well as requirements for sample preparation and fixturing. The potential for applications in the biomedical sector will be demonstrated.

Introduction

The continuous advancements in technology and medicine have lead to a variety of implantable devices that are currently used for the treatment of diseases and disorders of many patients worldwide. In addition to orthopedic implants and devices such as artificial joints and limbs, functional active implantable devices have gained significant importance. Pacemakers are the most widely used functionally active devices used in more than 2 million patients worldwide. Other devices that also benefit thousands of patients include implantable cardiac defibrillators for patients with an irregular heart rhythm, cochlear implants for hearing-impaired people, and neural stimulator devices for the treatment of seizures caused by epilepsy [1,2,3,].

The spectrum of potential applications for the next generation of implantable biomedical devices will be even broader. Driven by advances in micromechanical systems (MEMS) technology, nanotechnology, molecular diagnostics and drugs, new devices that are currently under development or that are already undergoing the extensive approval process, have the capabilities of monitoring, diagnosing and treating disease. Biosensors are one focus area of current biomedical device research. The range of applications reaches from the localized measurement of body temperature and pressure, for example in the brain, to monitoring of heart functions and blood sugar levels. Neural stimulator devices implanted in the brain are expected to effectively treat a variety of neurological disorders including Parkinson's disease [4,5].

Drug delivery implants that can locally effuse chemicals and drugs are offering more effective treatment with fewer side effects compared to administering the substances orally or through the blood stream. This device is based on a chip including an array of small fluid reservoirs that are sealed with a metal membrane. By applying an electric potential, the membrane can be dissolved and the drug is released [4]. Other concepts for implantable drug delivery systems are based on using micropumps and refillable reservoirs that can store the drug for several months [5].

For many applications, future implantable devices are designed to be long-term implants that are remotely powered and controlled. Their further development and deployment strongly relies on the availability of assembly processes that allow high integration, hermetic sealing and automated processing with superior quality. The development of new, biocompatible materials and manufacturing processes

that ensure long-lasting functionality and reliability is a critical issue.

Packaging of Implantable Devices

Significant challenges in the assembly and packaging of implantable devices result from the small size of the features, the heat sensitivity of integrated electronics and media such as drugs, the precision alignment required to hold tolerances, and the type of materials to be joined. In addition to metals, which are primarily titanium but also platinum, gold and stainless steel, non-metals play an even more important role based on their biocompatibility. Depending on the type of implant and its location in the human body, such as the brain or under the skin in the chest or neck area, implants are exposed to different environments and the encapsulation material must meet specific biocompatibility requirements.

The group of non-metal materials includes glass, sapphire, silicon and polymers. Based on their properties they are a preferred choice for specific device structures as well as for the encapsulation of the implants. Specifically glass provides good biocompatibility and is therefore suited for the encapsulation. The need to join dissimilar materials occurs if the encapsulation includes functional elements such as electrodes used for neural stimulation or modules with defined micro-pores for the effusion of drugs. Proven methods for localized joining and hermetic sealing of non-metals and dissimilar material combinations in particular, do not yet exist for implantable microsystems. Today, adhesives and soldering are widely used to join dissimilar materials, however there are drawbacks. Lack of long-term stability and biocompatibility are critical factors that limit their application for microsystems used in the biomedical sector. In addition, high heat input of soldering and the material shrinkage during curing of adhesives makes precision alignment difficult and automation very challenging.

Laser joining of biocompatible materials

Glass Welding

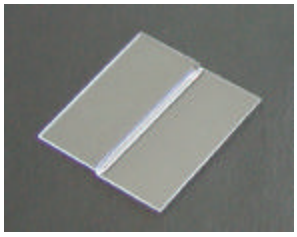
Laser welding of glass requires special considerations due to the thermal, mechanical and optical material properties. Typically, glass is characterized by low thermal conductivity, low to moderate tensile strength, and high brittleness. Localized heating with laser radiation causes high temperature gradients within the material facilitated by the low thermal conductivity and the fact that the CO_2 laser radiation is primarily absorbed at the glass surface and doesn't penetrate into the material. The local thermal expansion induces stress that often exceeds the material strength and leads to cracking. Similar residual stresses are induced during the cooling cycle and can also lead to cracking. In general, pre- and postheating are effective countermeasures to reduce temperature gradients during processing and therefore the risk of cracking, however their applicability to implantable microsystem is limited considering the heat sensitivity of electronics and other media inside the devices.

Laser welding of borosilicate glass (Pyrex 7740) was investigated using a pulsed CO_2 slab laser (ROFIN SCX-10). The laser provides a maximum output power of 100W at a duty cycle of 50%. It has a Gaussian beam profile and high beam quality ($K>0.8$) resulting in good focusibility and a small spot size. A scanning optic with 100mm focal length is used to focus and move the beam across the workpiece. Two measures were applied to reduce the risk of cracking without having to preheat the entire sample. First, the welds were performed with a defocused beam (focal position: 6 to 8mm) to reduce the temperature gradient within the heated zone. Second, a multiple scan strategy with high speed was selected to evenly heat the joint. Each scan covered the entire joint line, which is a valid approach considering the small size of the devices to be sealed.

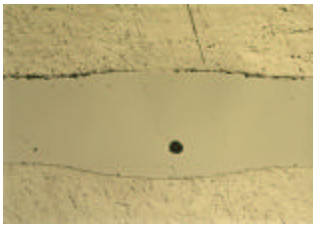
Welding tests were conducted on 0.5mm thick glass wafers with a size of 10mm x 10mm or 10mm x 20mm. A variety of joints were investigated including butt, penetration overlap, and fillet edge joints. Representative results are shown in figure 1. High quality crack-free welds were established with only minor defects. Only a few isolated pores occur that do not affect the joint strength or its hermeticity. Using multiple scanning, the energy input can be well adjusted to the individual requirements of the specific joint type. Most cross-section pictures show a very smooth transition between the joined glass wafers. The fillet edge joint is employed for specimens that are used for hermeticity testing. In this case, a thin (0.15 mm) top wafer is welded onto the 0.5mm thick square wafer, which has a hole in the center. The hermeticity is evaluated using a helium leak tester. To measure the amount of helium crossing the joint, helium is presented at one side of the specimen while a vacuum is applied to the other side. Tests revealed that the laser joints are hermetically sealed.

Good results in glass joining are achieved using the multiple scan strategy and the defocused laser beam. Both factors lead to a reduction of residual stresses in the joint and are important to prevent cracks. However, they also increase the interaction time between the material and the laser beam and result in

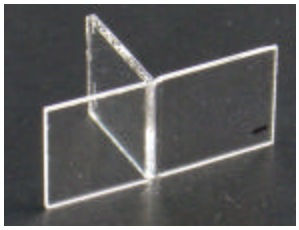
higher heat input into the parts. With regard to the application of this joining method to implantable devices, the process temperature is critical to avoid damage to the functional components inside the device as well as on the encapsulation's surface. Since glass joining with laser is based on melting the material, temperature levels along the joint line are expected to be close to or equal to the melting temperature of the material. Pyrex 7747 glass starts to soften at 821°C.



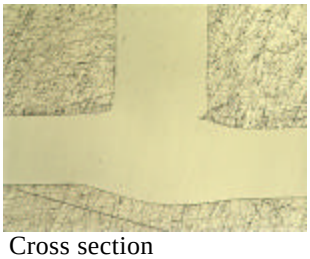
Butt joint
(weld length: 20mm)



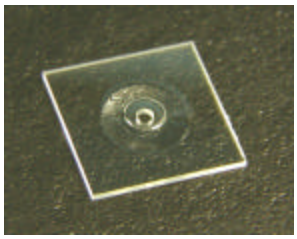
Cross section



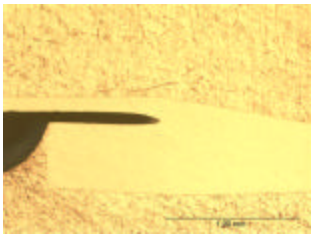
T-joint
(weld length: 10mm)



Cross section



Fillet edge weld (sample for leak testing)
Wafer size: 10mmx10mm



Cross section

Figure 1: Laser welded glass samples (material: Pyrex 7740)

To determine the actual temperature during laser welding in the joint line and in close proximity to it, measurements were conducted using a thermocouple. The 0.025mm wire was placed between two glass wafers that were joined with an overlap penetration weld. Figure 2 shows an example of two joined wafers with the thermocouple in-between. In this case, the measuring tip of the thermocouple is located 0.42mm from the center of the joint almost at the edge of the melt zone. Since the melting temperature of the thermocouple is significantly higher than the melting temperature of the Pyrex glass, the thermocouple is not damaged by the welding operation.

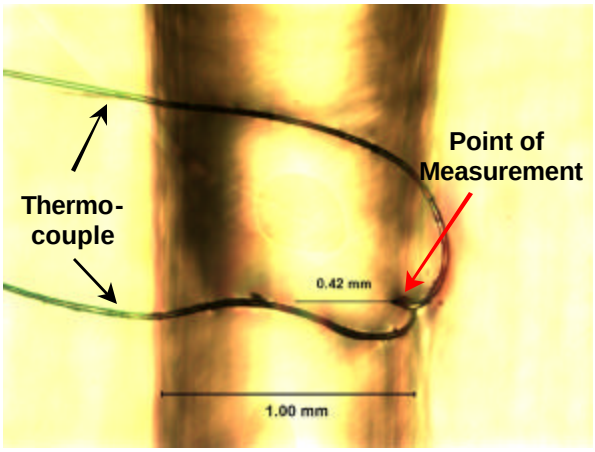


Figure 2: Penetration overlap weld with thermocouple

The results of these initial temperature measurements for 6 different locations in the weld and the area next to it are demonstrated in Figure 3. A temperature of almost 1000°C is reached close to the centerline of the approximately 1mm wide weld (distance: 0.28mm),. Within the melt zone, the temperature drops only slightly by less than 100°C from the center to the edge. The temperature decreases significantly in the heat-affected zone outside the melt zone.

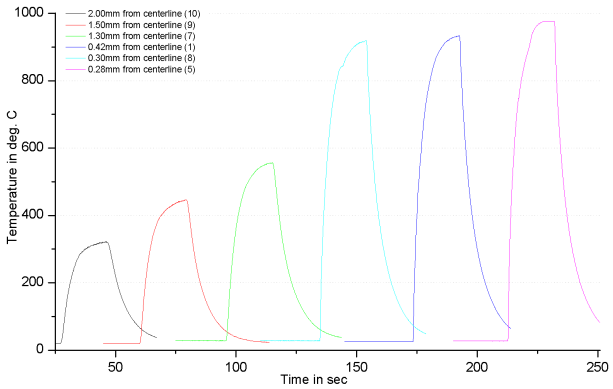


Figure 3: Maximum temperature for different distances from weld centerline

At a distance of 2.0mm from the centerline, the temperature during welding reaches a maximum of only 300°C. This temperature is still too high to be tolerated by most functional components of

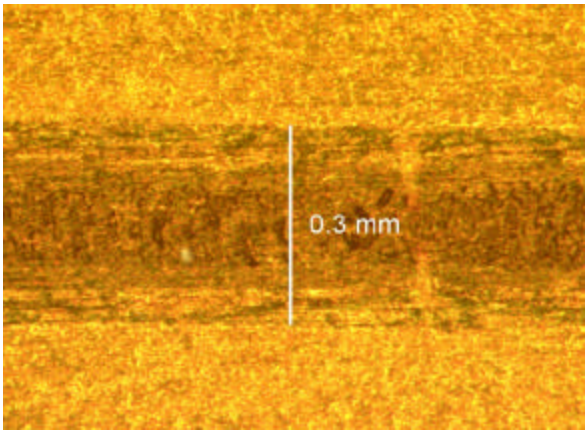
implantable devices. However, the weld was conducted on two 0.5mm thick wafers that require high energy input to be joined with a penetration overlap weld. Future work will focus on characterizing other joint configurations with thinner material regarding their temperature profile during welding. The challenge lies in the appropriate placement of the measuring tip within the joint or in the heat-affected zone. In general, all measurements show very plausible results with regard to the progression of the curves for the heating and cooling cycle and to the temperature level confirming the appropriateness of the measuring method used.

Laser Bonding of Polymers to Metallic Substrates

Biocompatible polymers that are used for a variety of medical applications such as Polyurethane (PU, Pellethane®) and polyimide (PI, Imidex®) were investigated regarding their joining feasibility to metals. Titanium and stainless steel (300series) were selected as metallic bonding partners because both provide certain levels of biocompatibility for specific implant applications. Two laser sources were applied to bond the dissimilar materials: a 30W fiber coupled diode laser (DIOWELD) with a wavelength of 808nm and a 25W fiber laser (JDS IFM25) emitting radiation at 1100nm. Compared to the diode laser, which provides a minimum spot size of 0.8mm, the fiber laser has superior focusing capability with a minimum spot size in the range of 20µm. Therefore, the fiber laser is better suited for applications that have small features and require very narrow bond lines.

The transmission joining principle was selected for the dissimilar material combinations since the selected polymers exhibit sufficient transmissivity for IR laser radiation. Acceptable bonding results on stainless steel and on titanium were achieved for both combinations. A representative sample of bonding of a 0.177mm thick polyimide film to a 0.05mm thick titanium film with the fiber laser is shown in Figure 4. The fiber laser beam was defocused to a diameter of approximately 0.13mm to lower the intensity in the center of the laser spot and to avoid overheating, degeneration and burning of the polyimide. In visual inspection, the overall bond width amounts to approximately 0.3mm, which is noticeably larger than the actual beam size. Possible reasons for the difference are scattering of the radiation in the polyimide film and the heat dissipation during processing. A darker zone is observed in the center of the bond line, indicating the differences in the energy input across the bond based on the beam intensity profile. In addition, there is a slight widening of the bond width from start to end that is primarily based on the pre-heating effect of the material during processing. This effect is less distinct at higher speeds.

Laser bonding with the fiber laser was investigated for power levels between 1 and 4W and speeds up to 3000mm/min. As shown in the process diagram (Figure 4), good bonds are achieved for a reasonably wide speed range at a constant power level. Therefore, the bond process appears to be forgiving of small deviations from the fixed parameter settings. Laser power levels were only investigated up to 4W. At this power level, good bonds are achieved at high speeds up to 2000mm/min.



Bond line (surface picture)
Power: 2.2W
Speed: 100mm/min

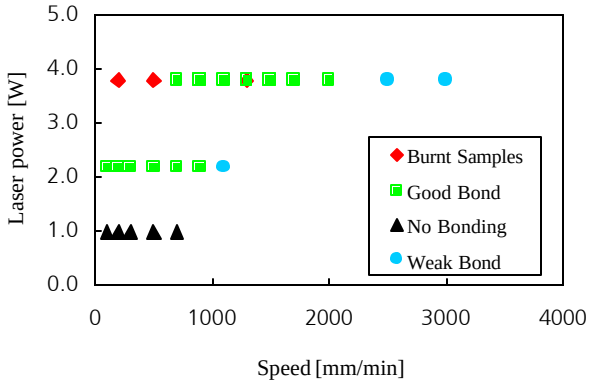


Figure 4: Polyimide-to-titanium bonding with fiber laser

The characterization of the joint quality for polymer to metal joints is part of the ongoing program to qualify these laser bonds for biomedical applications. The investigation includes: tensile strength measurements, microscopic analysis and x-ray photoelectron spectroscopy (XPS) of the bond interface and fractured

joint surfaces, hermeticity testing and degradation studies in physiological solutions to evaluate biocompatibility.

Tensile testing is performed using a multi-axis micro-tester that applies a controlled load to the specimens and pulls the samples until the bond fails. The set-up is shown in figure 5. The specimens used for tensile testing consist of two 10mm x 30mm pieces that are bonded at the end in an overlap configuration. The length of the bond line is 6.5mm, therefore the beginning and the end of the line are on the sample and any effects resulting from passing over the edge of the sample, such as heat accumulation are avoided.

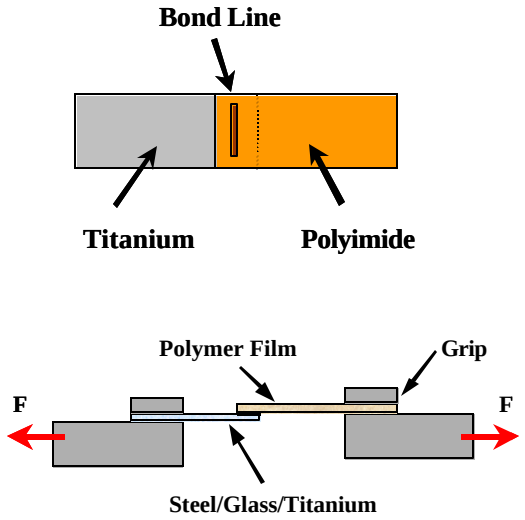


Figure 5: 6-axis microtester for tensile/fatigue testing

Thorough tensile testing was conducted for polyimide-to-titanium joints including an investigation on the effect of different laser parameter settings on the bond strength. Between 6 and 10 specimens were tested for 8 different parameter settings. The results are summarized in Figure 6 and demonstrate that the bond strength increases with lower scanning speeds. However, once a sufficient level is reached, the laser power has only minimal influence on the bond strength. For the investigated parameter ranges, the maximum failure load of 12.4N/mm^2 is achieved using 3.1W power and a scanning speed of 250mm/min.

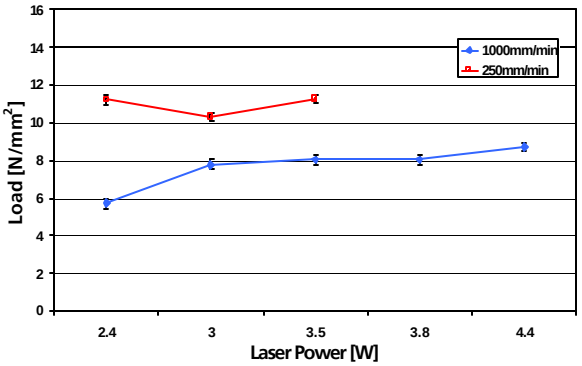


Figure 6: Failure load of polyimide-to-titanium bonds

In comparison, polyimide-to stainless steel bonds that were achieved using 2.7W power and a speed of 200mm/min only reach an average maximum failure load of 8.6N/mm^2. The inspection of the failed samples shows polyimide traces on both metal substrates in the center of the previous bond line. This indicates that the shear failure started at the edge of the bond at the initial load levels and was followed by the failure of the adherent polyimide at the successive load levels. It also suggests that the bond strength is higher in the center of the bond line.

XPS studies on the polyimide/titanium interface were conducted to further investigate the chemistry of the joint formation and research the established bond mechanism. XPS conducted on the failed bond surfaces revealed no presence of titanium on the polyimide side, indicating that the separation occurs directly at the interface or within the polyimide film. In contrast, spectra taken from the titanium surface showed the presence of nitrogen that can only originate from polyimide residue sticking to the surface. High-resolution XPS spectra taken of the polyimide residue on the titanium foil after facture are shown in Figure 7. An argon ion beam was used to sputter the surface and obtain in-depth spectra. While the spectrum taken at the polyimide residue surface shows the presence of carbonyl (>C=O) groups, such groups are not observed in spectra taken from areas that are closer to the polyimide/titanium interface. It is assumed that the double bond between oxygen and carbon is dissolved and that chemical bonds of the type Ti-O-C exist in a very thin interface with the thickness in the range of 10nm. In addition, the spectrum PI/TI t_{sp}=120s also

shows evidence of the existence of titanium-carbon chemical bonds. Both chemical bonds are expected to cause the relatively strong joint between these dissimilar materials.

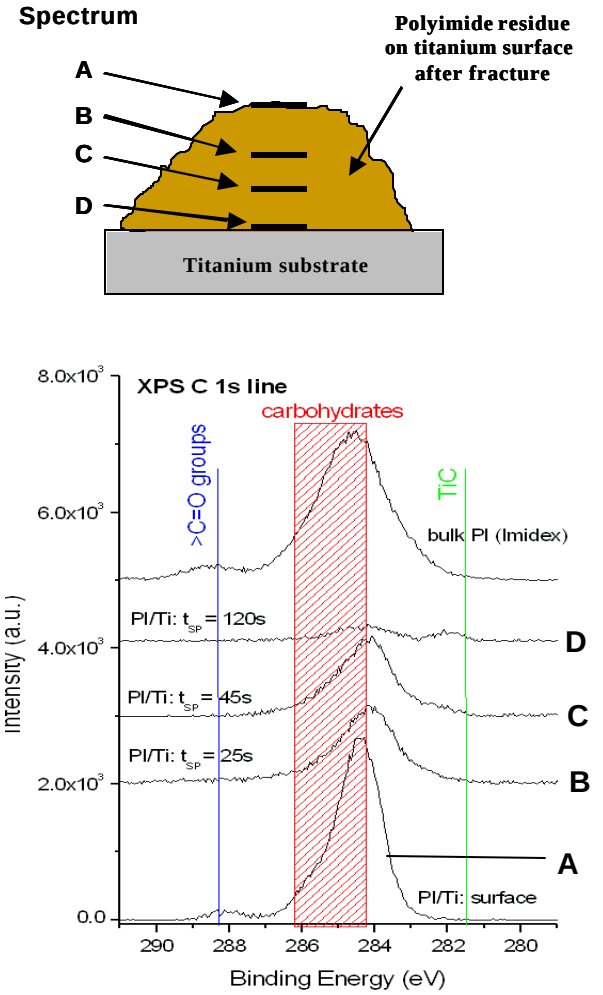
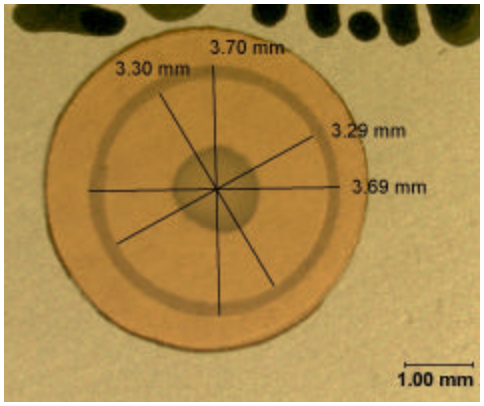


Figure 7: XPS spectra of the polyimide-titanium bond interface

Hermeticity tests were performed on special coupons that required a circular bond line. The coupons consist of a polyimide disc with a diameter of 5.0mm bonded to a 10mm x 10mm x 0.5mm titanium coupon with a center hole with a diameter of 1.0mm. A very consistent bond line without defects, especially at the start and end, is needed to achieve a leak-tight bond. Figure 8 shows characteristic results achieved with the fiber laser. Helium leak tests performed on these samples showed similar leak rates for the laser bonded sample and the polyimide film itself.



Power: 4.2 W
Speed: 100 mm/min

Figure 8: Circular bonds of polyimide-to-titanium on leak test samples

Conclusions

Next generation implantable microsystems that are currently under development will increasingly involve non-metallic encapsulation materials due to functionality and biocompatibility requirements. The improvements in the design and functionality of these microsystems require innovative packaging solutions. New, reliable and biocompatible joining methods for non-metallic materials and dissimiliar material combinations are needed that can be applied on a micro-scale. The feasibility of localized laser joining for a variety of dissimilar and non-metallic materials has been demonstrated. The results achieved in glass welding and bonding of polymers to metals show some promising onsets to address these joining challenges. Future work will focus on the further optimization of the laser bonding processes and the characterization of the achieved bond quality.

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