MASTER THESIS

SENSORIZATION OF LOWER LIMB SOCKET FOR REAL-TIME PRESSURE MONITORING

Roy Fimbres Ramirez

FACULTY OF ENGINEERING TECHNOLOGY
DEPARTMENT OF BIOMECHANICAL ENGINEERING

EXAMINATION COMMITTEE
Prof. Gabriëlle Tuijthof
Dr. Izadyar Tamadon
Dr. Islam S. M. Khalil

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Abstract— Lower limb amputees patients experience an extraordinarily high incidence of skin lesions and eventual prosthesis abandonment. Common issues are pain and discomfort, which can be attributed to high local pressures. Patients face volume fluctuations both in the short and long term, which can hinder proper prosthetic use. To address this issue, a sensory system is proposed that focuses on key areas required monitoring by employing inductance properties of coils (Fig. 1). Here in this report, three conical Spring Sensors (SS) are explored and evaluated through compression tests, demonstrating a repeatable response to changes in length. The linear models for the spring sensors SS1, SS2 and SS3 exhibited mean relative errors of 0.2355, -0.0531, -0.0057, respectively. Moreover, SS2 and SS3 can estimate the force with a mean relative error of -0.0014 and -0.1156, and RMSE and Standard deviation of 0.00483±0.0049, 0.0101±0.0102 uH respectively. These characteristics, combined with their low cost and compact size, make soft sensors ideal candidates for pressure sensing. Moreover, the proposed sampling areas for the sensing array offer an alternative to address the inherent challenge of monitoring the residual limb. This paper provides a proof of concept for the proposed model and inductance sensors as a viable alternative.

I. Introduction

The World Health Organization (WHO) claims that only 1 out of every 10 people in need can access medical assistive devices [1]. Even when patients have access to prosthetic devices, rejection is a common occurrence. Researchers proved that the main factors for rejection of prosthetic devices are pain, discomfort, and high stresses which can lead to Pressure Ulcers (PUs), which can lead to further amputation or even result in life-threatening injuries [1], [2], [3], [4]. The importance of a proper socket fit can be highlighted by the vast number of studies aiming to measure socket-fit and comfort success [3], [5], [6]. Consequences of improper socket-fit include, but are not limited to, gait deviations, long-term musculoskeletal degradation, skin irritation and breakdown, heat, and sweating, as well as dermatitis and infections [3], [7], [8]. Skin lesions are commonly observed in lower limb amputees with a 60-82% incidence rate, and a subsequent rate of device use abandonment between 25-57% [2], [5]. Previous studies showed challenging rehabilitation for transfemoral amputees, and an underlying correlation between high-level amputation, which in the case of lower limb refers to above the knee amputation, and prosthetic abandonment [7].

The residual limb experiences both short and long-term volume fluctuations, due to factors such as fluid level fluctuations, and the development of compensation strategies. To account for the function of the missing limb, despite muscles in the residual limb remaining fully functional the recruitment process is altered. These strategies lead to instances of localized atrophy and hypertrophy [4], [5], [6]. Studies have found a maximum increment of 7% and a maximum decrement of 11% over the volume, it was also observed that a 3-5% fluctuation is enough to generate a high degree of discomfort [5], [6]. The ideal socket should be able to manage these fluctuations. Current solutions are based on clamp systems or lacing that must be adjusted manually [5]. This becomes a larger issue as a significant portion of lower limb amputees live with other diseases such as vascular issues, diabetes, and sensory impairments associated with old age, the ability, thus relying on the user feedback is compromised [3], [5], [8]. A system to monitor the residual limb would benefit all lower limb patients. It must be highlighted that diabetic patients would gain the most benefit as the consequences of ill-fitting prosthetics are usually worse for these patients [3]. In current practice, prosthetic manufacturing relies heavily on empirical methods, which are labor-intensive, material-wasteful, and highly dependent on the prosthetist's expertise, with limited input from the amputee [9].

Following traditional designs, an adjustable socket seems to be the best viable option for these patients, as it adapts to evenly distribute the pressure along the residual limb, thus relieving pressure from sensitive areas ultimately enhancing the user’s comfort [7]. Continuous socket volume adjustments have been proven as an effective tool for managing residual limb volume [7]. In recent years, a growing interest in developing user friendly tools to sense and monitor the inside of prosthetic sockets has become evident. Studies have been conducted using 3D printed sensors, in liners with sensors, and detachable inserts with sensing instruments [3]. The development of conventional and flexible electronic devices to monitor physical activities within the body has also been subject to interest in recent years [10]. As technology advances, the possibilities of incorporating cheap and reliable sensors keep growing. One of the main challenges for limb monitoring is the current absence of feasible inner socket sensors capable of monitoring the interface between the skin and the liner or socket [7].

Technical challenges and practical issues regarding sensor technologies make recording pressures across the whole residual limb virtually impossible. Therefore, the location of sensors is considered a critical factor, as it can interfere with the socket [7]. Previous works in the literature have made attempts to monitor the pressure inside of the socket, however, said efforts resulted in systems with bulky sensors, wires, cables, and/or required holes in the socket, thus affecting the environment within the socket and compromising the value of the measurements. Consequently, the measurement instruments were deemed unfit for everyday use [3].

The purpose of the proposed system is to actively monitor the pressure within the lower limb socket as a mean of preventing issues such as skin lesions and infections, with the end goal of alleviating pain, increasing patient comfort and reducing negative outcomes like prosthetic abandonment and subsequent amputations.

DESIGN REQUIREMENTS

There are four main types of sensors used for pressure measurements in the socket: strain gages, piezoresistive, capacitive, and optical sensors [6], [7], [9]. In general, these sensors have some inherent challenges such as difficulty with calibration, accuracy and hysteresis, additionally there have been recorded issues with sensor movement, crosstalk between sensors and interference with the residual limb and socket interaction [9]. Some additional issues might present themselves depending on the type of sensor. Strain gauge sensors stiffness can lead to stiffness mismatch with the surrounding tissue and liner material. This mismatch can result in stress concentration at the sensor edges, causing local tension in the tissue of the residual limb [6], [11]. Optical sensors have a particularly high risk of being damaged by use therefore their use is limited [5]. Finally, piezoresistive and capacitive sensors are the most common in literature. These last sensors might exhibit problems such as nonlinearity over the full pressure range and parasitic noise [5], [12].

Therefore, the proposed system should be able to overcome the limitations of commercial options. The system should exhibit minimal hysteresis, linearity and good accuracy and repeatability with minimal crosstalk among sensors. Additionally, the sensors should be compact enough to fit into the socket/in-liner without compromising comfortability. As a reference the thickness of the widest inliner from Wilowwood, at 9mm, is taken as a point of reference [13]. Due to the challenges of recording pressures around the whole residual limb the system should place sensors in key locations to assure proper monitoring.

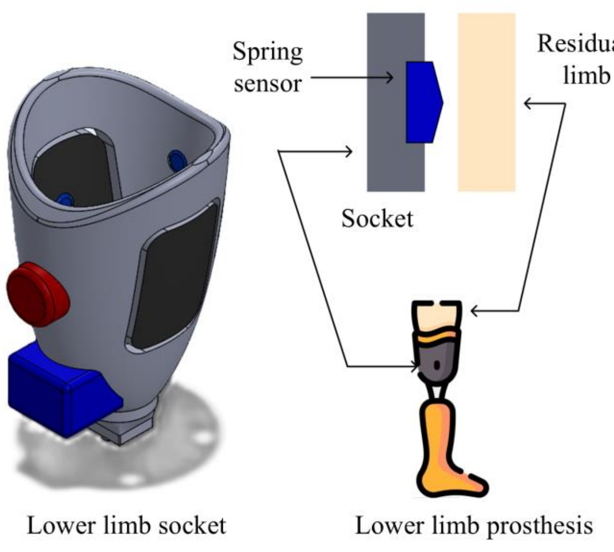

Fig 1. System Overview (Lower limb prosthesis icon retrieved from Flaticon.com)

The type of sensors is a crucial aspect, in this article instead of focusing on pre-existing technologies for in-socket pressure measurements, an alternative is explored. Flexible pressure sensors have been applied for wearable health devices, prosthesis, and human–computer interaction applications, due to their flexibility, cost, and sensitivity [14]. Soft inductive sensors have a reliable gauge factor and excellent electromagnetic compatibility. Soft sensors have better repeatability and resistance when compared to traditional sensors, because the signal is generated from a distance change rather than from an unstable change. However soft sensors might exhibit hysteresis which has an impact over their dynamic precision [15]. In recent years, spring-based inductors have been explored and studied as an alternative for biomedical applications. Inductive sensors have been proposed as a valuable alternative for strain and tactile sensing, due to their low hysteresis, stretchability, and signal repeatability. Studies highlighted that sensing through these devices can be measured by either the elongation, bending or a combination of both [16]. The use of conical springs to monitor inductance changes represents an opportunity to monitor and record changes in volume and pressure with a low-cost, additionally the spring designs allow them to become flat when compressed this attribute is especially useful in settings like the socket where there is limited space for sensors, and current solutions are all too bulky. Experiments performed by Xing et al., show favorable results, showcasing high repeatability rate with less than a 0.1% repeatability error [15]. In a similar study by Sahu et al. similar results were achieved. The system exhibited favorable characteristics, low hysteresis at 0.1%, a high precision (in the order of 0.14%), and a high accuracy of 0.9%, all of which outperform common resistive and capacitive alternatives [16].

Another crucial aspect to consider for a functional system is the positioning of the sensors. Fereira et al. proposed a system considering the Anterior-Posterior (AP) and Medial-Lateral (ML) planes. The AP plane was established in relation to the sagittal plane, while the ML was established perpendicular to the AP plane, aligned with the frontal plane. Both the AP and ML planes were recognized by prosthetists as areas designated to bear load [17]. Literature highlights Anterior proximal, Anterior distal, and Posterior distal positions inside the socket, to be positions of interest, as they generate the maximum stresses during gait cycle [18], [19]. Some studies focused on Medial and Lateral profiles to record the in-socket pressures [20], [21]. Whetersby et al. developed an arrangement compatible with RevoFit kits, by arranging the sensors in six positions namely Anterior proximal, Anterior medial, Anterior distal, Posterior medial, Posterior distal and Posterior lateral, in this case the socket had adjustable panels, placed over commonly used sites as used by RevoFit [22]. The proposed arrangement assumes a flexible socket with adjustable panels located at the medial and lateral profiles, and a mechanism to adjust them at the anterior profile. Sensors are placed at the anterior proximal, anterior medial, and anterior distal positions as these areas have been identified in several studies as key pressure points [7], [18], [19], [22]. Additional sensors would be positioned at the medial and lateral locations to track changes in pressure resulting from tightening or loosening the adjustable panels of the socket. As these parts are adjusted there is an expected change in inductance resulting from said adjustment, which should be monitored. The proposed arrangement covers the main key regions presented by Ko and could provide an accurate description of in socket pressures, as the patient goes through the day [7]. Fig. 2 shows an overview of different studies and the proposed locations for sensor placement [18], [19], [20], [21], [22]. This selection considers planes as described by Fereira et al. [17].

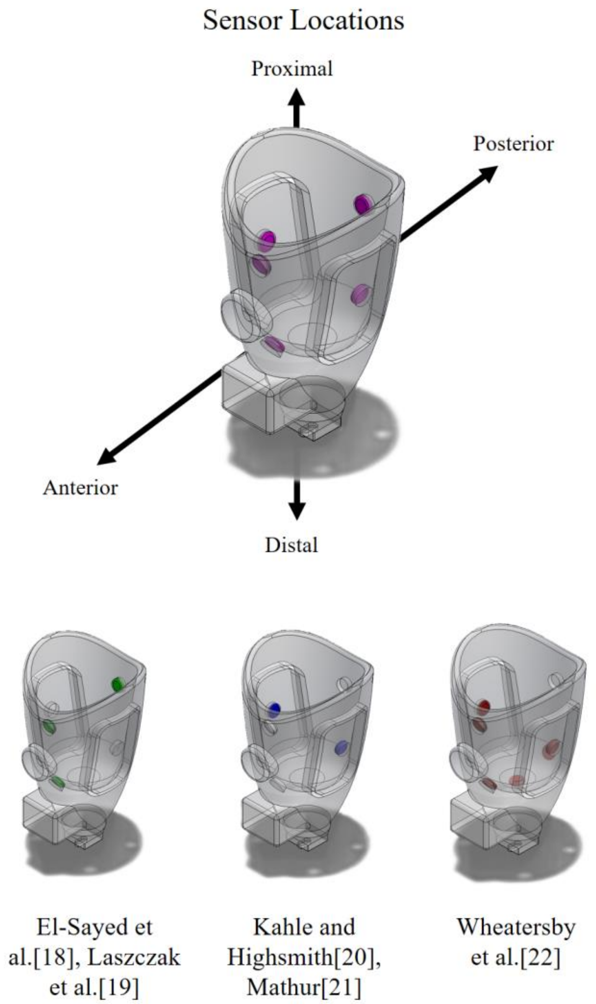

Fig 2. Sensor placements , Top image portraits the proposed sensor placement. Lower images from left to right showcase the placement used by other researchers. Anterior proximal, Anterior distal and Posterior proximal by El-Sayed and Laszczak. Medial and Lateral by Kahle and Highsmith and Mathur.And at last Anterior proximal, Anterior distal, Posterior proximal, Medial and Lateral by Wheatersby.

A system utilizing inductance-based spring sensors is proposed, integrating an arrangement of soft sensors at the identified key locations, to enable low-cost pressure monitoring by measuring the inductance of the coils. This work aims to provide proof of concept for using conical springs to monitor pressure within a socket. To achieve this, the changes in inductance were tested at various lengths, and the optimal sensor placement was carefully analyzed. The compact size of these sensors offers a significant advantage over existing solutions, making the system more comfortable and compact. The detailed study of sensor characteristics and their arrangement highlights the potential of this system for effective pressure monitoring.

The article is organized as follows. Section II provides a brief introduction to inductance functional principle and outlines the utilized methodology for the observation of inductance at different lengths and at different pressures. Section III presents the experimental results and corresponding equations. Section IV discusses the benefits and limitations and describes possibilities for future work and applications. At last section V draws the overall conclusions and highlights the impact of the proposed system.

II. Materials and Methods

A. Materials

In principle, when current flows through a coil or inductor, it generates an electromagnetic field. Any changes in the current are mirrored by the field. These variations in the field produce a voltage within the coil that opposes the changes in current, a phenomenon known as inductance (L). Inductance is inversely proportional to the core length and directly proportional to the cross-sectional area [23].

The inductance of the springs was measured using a commercially available circuit board (LDC1614EVM) along with the manufacturer’s software. Three models of conical compression springs were acquired from Sodemann Industriele-Veren, and stainless steel 302 springs were specifically chosen to evaluate inductance variations. Table 1 presents the characteristics of these springs, all of which share the same unloaded length of 9.53 mm, ensuring a consistent testing method. Previous studies indicated that the forces measured were approximately 26N, using sensors with a 30N measuring range[18], [24]. Based on this, the springs were selected to endure at least 26N, with the maximum load from the chosen components being 52.4N. While SS1 can support a maximum load of 24.95N, it was deemed acceptable with a 10% tolerance.

To accurately characterize the sensors inductance, two setups were developed: one focused on inductance changes due to length variations, and the other on how varying pressures affected inductance. By dividing the tests into these two setups, a better understanding of the sensor and potential performance in the socket is achieved.

The first setup was built to accurately test the springs length, a setup was 3D printed with PLA, and plastic screws and nuts were added. The system can be observed in Fig. 3 . In this setup, a M3 plastic screw was added at the base and secured with LOCOTITE super glue, and subsequently secured with a PLA ring that was heat welded to the base. The length adjuster (blue component) encloses a matching nut for the screw, this allows loading and unloading of the screw. The setup also has a top ring which keeps the adjuster in a straight and upwards position preventing the system to tilt and, therefore, minimizing errors due to bending of the springs. The setup was designed in such a way that every turn of the adjuster translates into a 0.5mm length change. The springs were placed in the same positions, and the circuit board was connected to terminals for an improved contact with the springs.

The second setup consisted in 3 pieces of drywood, two tubes measuring 25cm in length and 20mm in diameter, and 4 3D printed pieces, 2 base guides and 2 bed attachments. Springs were placed into 3D printed covers made from Thermoplastic polyurethane (TPU), and inductance was tested with different weights. This setup was built to progressively add weight to the sensor and record the resulting inductance.

Table 1. specification of the sensor springs.

Specification	Spring Sensor1 (SS1)	Spring Sensor2 (SS2)	Spring Sensor3 (SS3)
Part No.	A600-040-037	A600-045-037	A720-055-037
d- wire diameter (mm)	1.02	1.14	1.4
L_0- unloaded length (mm)	9.53	9.53	9.53
L_{max}- unloaded length (mm)	2.04	2.28	2.8
S_{max}- max travel (mm)	7.49	7.25	6.73
F_{max}- max load (N)	24.95	41.14	52.4
D1- base diameter (mm)	15.24	15.24	18.29
D2- top diameter (mm)	7.92	6.35	7.92
Weight (g)	2.14	2.73	4.72

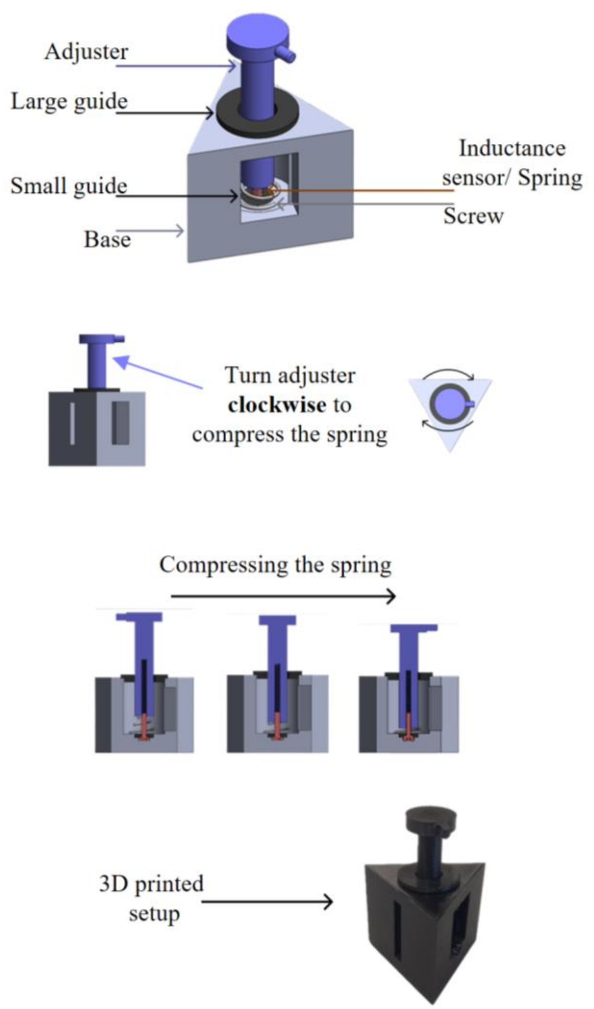

Fig 3. 3D Model experimental setup.

B. Methods

In both setups inductance data was then arranged in excel spreadsheets by tests and processed by MATLAB 24.1.0.2653294 (R2024a) Update 5. Consulted research collected data employing an operating bandwidth of 0-35 Hz [18], preliminary testing to determine the optimal bandwidth configuration was performed. The LDC1614EVM GUI (Graphical User Interface) allows for the adjustment of the Input digital filter and the Reference clock source which were set at 33MHz and 40Mhz respectively, with these settings the calculated sensor data oscillates around 21MHz.

For the first setup, Inductance measurements were taken in 3 separate sets, first by adjusting the spring length in 1mm intervals, this was chosen as smaller changes in length complicated observations on inductance changes. Then 2 sets were performed by turning the adjuster twice (approx. 1mm) one set for loading and one set for unloading. The tests were conducted five times for each measurement and sensor to assess accuracy and repeatability. Then, the inverse Inductance-Length was tested to validate and ensure the repeatability of these results, comparing the calculated values to experimental values. For these tests the minimum, maximum and mean errors were also computed in MATLAB.

Tests with the second setup (Fig. 4) were performed with springs SS2 and SS3, by applying 0,2,4,6, and 7.25 Kg additional tests were performed with 2.5, and5 Kg. SS1 was excluded from this study because its previous performance indicated it would not be suitable for this test.

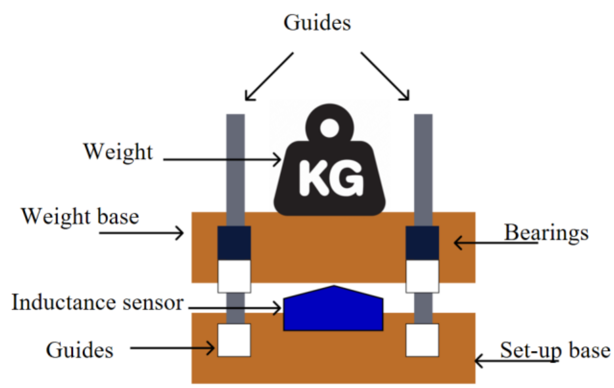

Fig 4. Experimental setup, inductance test with different weights.

III. Results

Results are arranged on Fig. 5. in columns by sensor and rows by test. First Inductance was tested at 1mm intervals, a Length-Inductance characterization and mean inductance (first rows and second row of Fig. 5.) were calculated with MATLAB, the results were processed to compute the following linear models:

SS1 $$L(S1) = .16357 + .00023X \quad (1)$$

SS2 $$L(S2) = .15197 + .0022X \quad (2)$$

SS3 $L(S3) = .15419 + .0021X$ (3)

In which L is inductance [uH] and X represents sensor length in [mm].

The linear models exhibited a standard deviation and a RMSE as seen in Table 2.

Table 2. Standard deviation and RMSE of Sensors SS1-3

Sensor	Standard deviation [uH]	RMSE [uH]
SS1	0.0018	0.00184
SS2	0.0027	0.002772
SS3	0.0025	0.00254

The linear model was subsequently compared to the inductance values gathered during loading and loading testing, as seen on the Length-Inductance fitting on linear model at third row of Fig. 5. For all spring sensors a linear fitting was calculated with MATLAB, for characterization, loading and unloading and there were no significant differences. Finally, tests were performed once more inversely, meaning that instead of setting the length and recording the inductance, the average inductance was set and the length recorded. Inductance-Length testing resulted in the graphs shown in the fourth row of Fig. 5. Additionally, for the Inductance-Length tests, the relative error, between the calculated and real lengths, was calculated. Relative error is calculated with equation (6) for all inductance points, the minimum, maximum and mean relative errors can be observed in Table 3.

Relative error $$RE = \frac{Experimental\ val. - Real\ val.}{Experimental\ val.} \quad (6)$$

Percentage error $$PE = RE\ x\ 100 \quad (7)$$

Table 3. Calculated minimum, maximum and mean inductance relative error.

Sensor	ERROR		
	Min	Max	Mean
SS1	-0.6763	4.1078	0.2355
SS2	-0.3549	0.2761	-0.0531
SS3	-0.6190	0.3117	-0.0057

Tests from the second setup also exhibit a correlation. As weight compresses the springs the inductance becomes lower, as observed in Fig. 6, this is in-line with what previous tests showed

SS2 $L(S2) = .21474 - .0028W$ (4)

SS3 $L(S3) = .21817 - .0065W$ (5)

In which L is inductance [uH] and W represents the weight on the sensor in [Kg].

With this second setup the linear models exhibited a standard deviation and a RMSE as seen in Table 4.

Table 4. Standard deviation and RMSE of Sensors SS2 and SS3.

Sensor	Standard deviation [uH]	RMSE [uH]
SS2	0.0049	0.00483
SS3	0.0102	0.0101

Relative error was calculated between calculated values and experimental values, the minimum, maximum and mean errors can be observed in Table 5.

Table 5. Calculated minimum, maximum and mean inductance relative error.

Sensor	ERROR		
	Min	Max	Mean
SS2	-0.0902	0.0647	-0.0014
SS3	-0.0622	0.2247	0.1156

Results from both experiments indicate excellent repeatability. For comparison, other sensors have reported repeatability error greater than 2% [15], [25], [26], [27], while inductance sensors have shown repeatability error close to 8% [28], [29], [30], [31], except for the works of Xing with 0.1% [15]. Therefore, in the present studies, repeatability error was expected to be in the range of 1-8%.

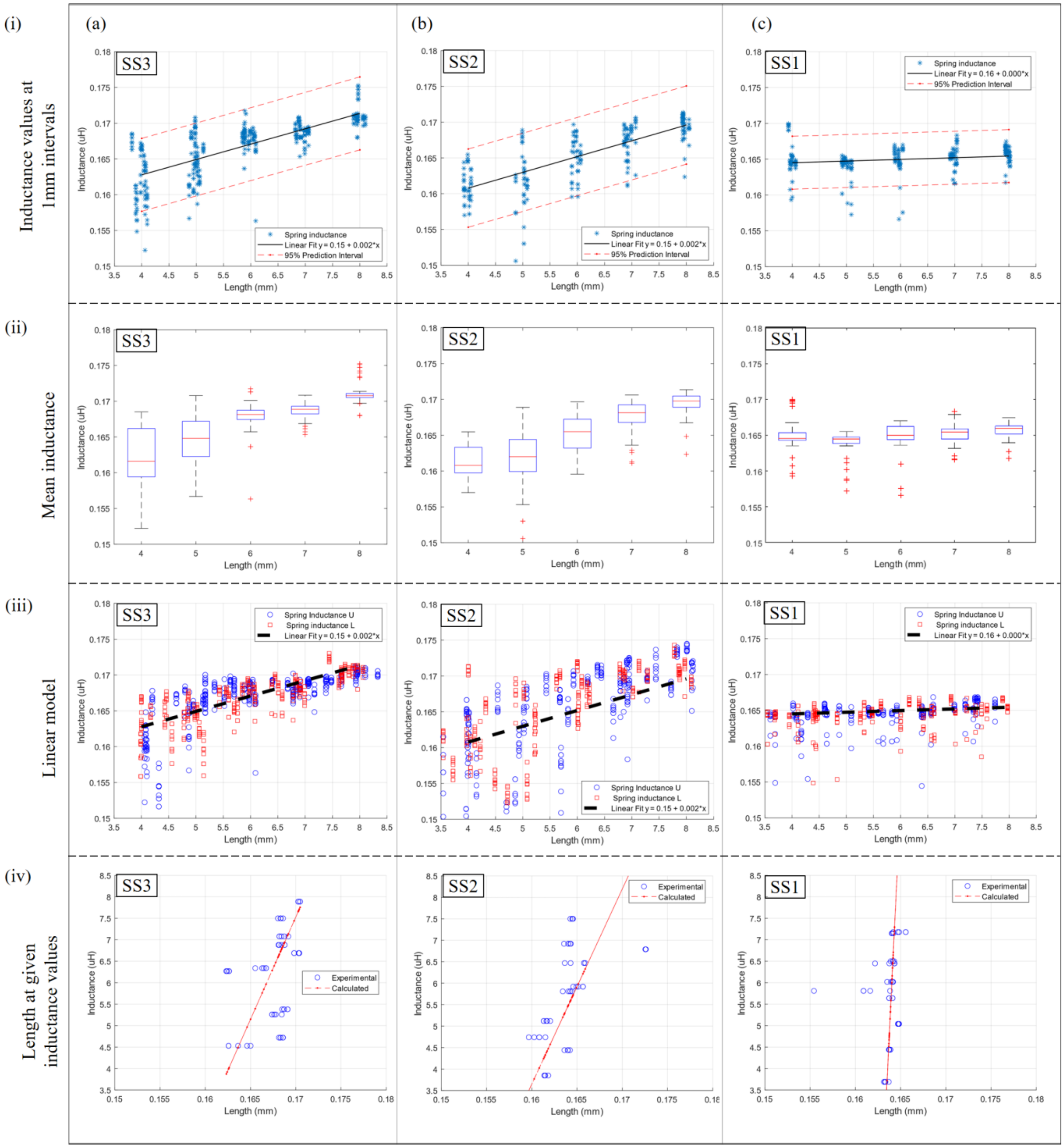

Fig 5. Experimental values, sensors SS3, SS2, and SS1 from left to right. From top to bottom (i) Inductance values at 1mm intervals, (ii) Mean inductance, (iii) Linear model, and (iv) Length at corresponding inductance values from the linear model.

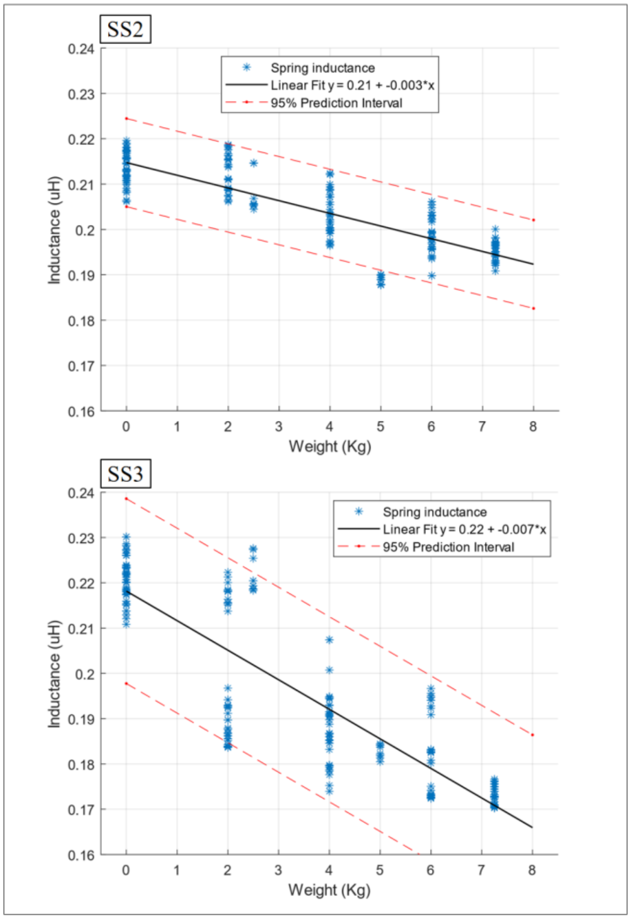

Fig 6. Experimental results 2, inductance of the conical springs under different weights.

IV. Discussion

The consulted literature proposes the use of soft sensors in medical applications to measure elongation and bending angle changes [15], [16], [32]. Table 5 shows a comparative overview of the proposed system. The proposed sensors have notable advantages in terms of repeatability and hysteresis when compared to other resistive, capacitive and, inductance sensors [15], [16], [30]. In these studies sensors performed with a mean error percentage of 7.9, however excluding the error percentage exhibited by SS1 this number decreases to 4.9. The experiments in this work were conducted with a sensor frequency of approximately 21 MHz, and the frequency parameters were the same for all sensors. However, the sensors may exhibit slight performance differences and behave differently under varying conditions.

Table 6. Performance comparative of proposed system and similar works.

Type of sensor	Resistive [15], [33], [34], [35], [36]	Capacitive [15], [25], [26], [27]	Inductive [15], [28], [29], [30], [31]	Present work
Repeatability error	Poor	>2%	0.1-7.8%	1-7.9%
Hysteresis	Poor	>0.8%	10%	ND

There is a noticeable linear relation between length and inductance in a conical spring, this suggests that using the inductance of springs to monitor pressure, socket fit could be improved. Although the results were not exactly as expected, they might improve with some adjustments. The linear fit of the spring sensor characterizations hints towards the possibility of implementing this kind of sensing mechanism as a viable alternative to modern commercial options. There were instances with discrepancies between the expected values and the experimental values. However, this can be explained by the arrangement of the setup, as inductances are quite small quantities (µH) slight changes can have a significant impact on the measured values and these last tests were performed in a different setup therefore cables and connections had different arrangements. One potential issue is that due to the inherent variance of the sensor the sensor can output similar values within a range, Hamaguchi et al. encountered the same issue with their research [32]. Following that issue, it can be highlighted that the system has the largest standard deviations when the spring is compressed the most (total length <5mm) these might indicate an adjacent challenge where the sensor tilts or touches itself generating interference with itself. The recorded data indicates a better resolution might be observed with springs capable of withstanding a higher maximum load. The sensors wire diameter might affect the inductance response as a SS2 and SS3 have a better resolution than SS1; wider wire diameter results in a lower number of coils for the spring. Other studies also pointed out that widening sensors resulted in enhanced sensitivity [32].

In the proposed device, sensors can be embedded into the socket, in a similar approach to the one used in Larsen's studies [37]. It is important to note that cable management might possess a challenge for integrating soft sensors into sockets as it significantly affected the inductance output in these experiments. For these studies, the cables were secured in place using tape. Finally, it can be mentioned that even though this study focused on providing proof of concept, added benefits such as improved thermal behavior and comfort might be achieved by implementing the proposed system with adjustable panels [5].

V. Conclusion

The present article, a system to monitor pressure within a lower limb socker through inductance changes in conical springs is proposed. For this system, the attributes of the sensor are just as important as the placement. Through literary research the optimal focal points were pinpointed and selected according to previous studies [7], [18], [19]. Given the characteristics of conical spring sensors they were deemed fit for this application.

This article proposed an inductance-based system as an alternative to monitor the in-socket pressure experienced by lower limb amputees. The system employs inductive sensors as an alternative to current sensing solutions. A sensor arrangement for transfemoral lower limb amputees is presented. The described socket makes use of inductance sensors and adjustable panels. To assess the viability of the concept three conical spring sensors made from stainless steel were tested. Tests focused on the length and the weight the sensor was exposed to. Soft sensors exhibited great repeatability and no observable hysteresis. The proposed system is cheap, reliable and more compact than current alternatives, therefore it is considered as an alternative worth exploring in further depth.

Integrating conical springs into sockets for monitorization of lower limb pressure might be a viable alternative, given their small size, compressibility and performance. The concept presented can be further improved and developed for other prosthetic and orthopedic devices with minimal modifications. This kind of spring has the singular advantage of compressing into a very thin space. Due to their characteristics and relatively low cost the use of soft sensors is a promising substitute for current options. However, the proposed sensor locations still need to be further studied and tested in the future. For future studies, it is recommended to use wider springs, or different materials capable of withstanding larger maximum loads, as this might help improve the resolution. Also, if possible, adding thermal sensors is advised, as temperature is also linked to negative outcomes regarding limb fluctuations and skin issues [5], [6].

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VIII. Appendix

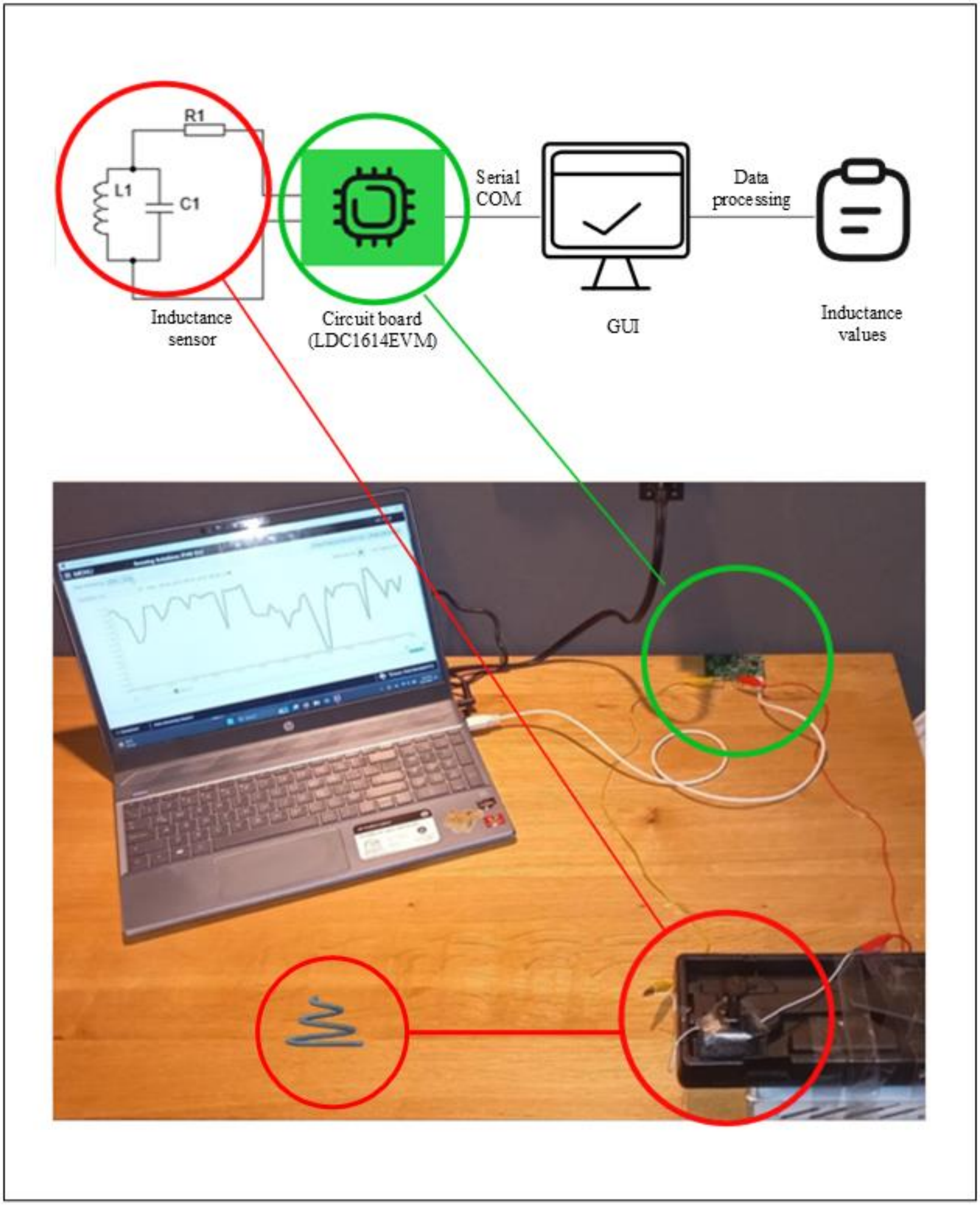

Fig 7 Schematic and wiring 1st inductance measurement setup. Top part shows a schematic representation of the inductance sensor (highlighted in red), connected to a circuit board (highlighted in green) to measure inductance this is connected to a computer through a serial communication cable, data is read through the Graphical User Interface, and stored for processing. In the bottom part the setup can be observed.

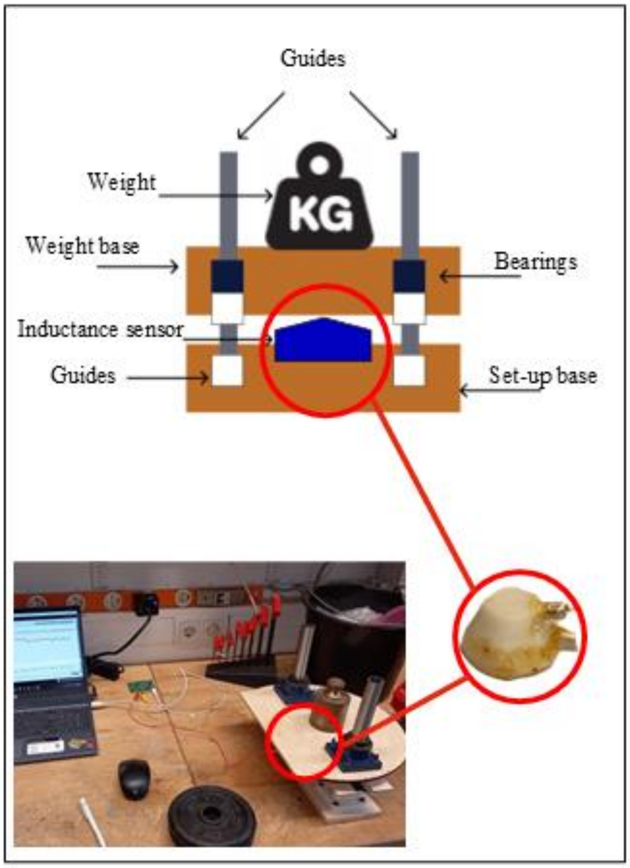

Fig 8 Inductance setup 2. In this setup the sensor (highlighted in red) connects to a circuit board and then to a computer just as in Fig. 7. The sensor is enclosed by a TPU cover and set in place where pressure is applied by adding weight progressively.