Monitoring diver kinematics with dielectric elastomer sensors

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ABSTRACT

Scuba diving is performed by millions of people for the recreation industry. There is a growing need however, to improve the safety and comfort of divers. The Divers Alert Network has reported on average 90 fatalities per year since 1980. Furthermore an estimated 1000 divers require recompression treatment for dive-related injuries every year. This study has investigated the feasibility of monitoring diver motion as a potential solution. Kinematic information can improve buoyancy control, provide a platform for gesture communication, and improve finning technique.

As kicking is one of the primary motions of a diver it is the focus of this investigation. Four dielectric elastomer strain sensors were fabricated and coupled to a wetsuit. The first two sensors were attached over the knee joints, with the remaining two attached between the pelvis and thigh. Experiments were completed with the participant wearing the suit in a freshwater swimming pool. A floating data acquisition unit monitored the sensors and transmitted data packets to a nearby computer for real-time processing. A GoPro Hero 4 silver edition was used to capture the experiments and provide a means of post-validation. The accuracy of joint angle measurements was assessed by examining GoPro footage in the image processing software, ImageJ.

This paper applies dielectric elastomer sensor technology to monitoring the leg motion of a diver. The experiment set-up and results are presented as well as a discussion into the implementation of strain sensors in human motion measurement.

Keywords: dielectric elastomer, underwater sensing, electroactive polymer, kinematic monitoring

1. INTRODUCTION

Diving has expanded globally both recreationally and commercially.[1] By 1999 there was almost 9 million certified divers in the USA alone.[2] Today, in addition to recreational purposes, diving is essential to military and police operations,[3] the oil and gas industry,[4] scientific research,[5,6] and sewer and water infrastructure maintenance.[7] However, as with every activity, diving poses its own risks. The Divers Alert Network has reported on average 90 fatalities per year since 1980.[2] Furthermore an estimated 1000 divers require recompression treatment for dive-related injuries every year.[8]

There is a growing interest in improving diver safety and comfort, which has lead to the development of several technologies. Two examples include Logosease, which allows face to face communication between two buddies, and Diverguard, which monitors air pressure and raises an alarm if the diver stops breathing. This study has investigated the feasibility of monitoring diver motion through a wetsuit with integrated strain sensors. Kinematic monitoring can potentially aid with buoyancy training, improve finning technique, and provides a tool for assessing fitness and health levels. As leg movement is one of the primary motions performed by a diver it is the focus of this investigation. More specifically the aim can be split into two goals: to record the count and timing of kicks and to measure leg joint angles.

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A wetsuit with integrated sensors is presented in Section 2. The experiment procedure and results are outlined in Section 3 and 4 respectively. A discussion on the investigation and implementation of strain sensors in an application is in Section 5. Finally an outlook is given in Section 6.

2. SENSING WETSUIT

Four strain sensors were integrated into the leg portion of the wetsuit: two between the hip and thigh, and two over the knee joints. 3M dual lock fasteners were used to attach the sensors to the garment. A floating data acquisition unit monitored the sensors and transmitted data packets to a nearby computer over Bluetooth. Sensors were connected to the data acquisition unit via 0.8 mm high tensile strength coaxial cable. The experiment and garment set-up are summarised in Figure 1.

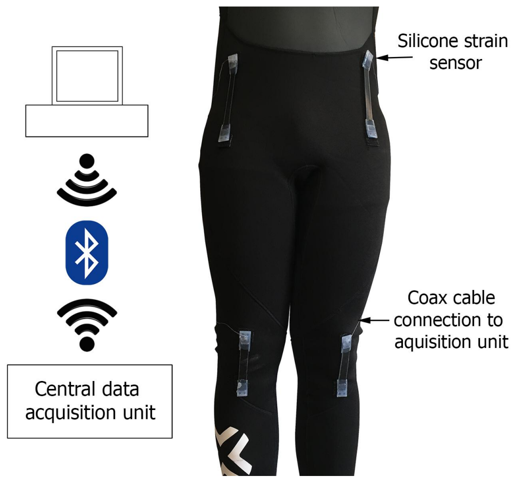

Figure 1. Wetsuit with sensors attached and communication summary. Electronics were housed in an IP67 rated container floating on the surface.

2.1 Sensor Characterisation

Soft and flexible dielectric elastomer (DE) sensors were used for strain measurement. The DE sensors comprise three elastic electrode layers sandwiching compliant silicone dielectrics. The electrodes are a carbon-doped silicone mixture. The sensor is entirely encapsulated in silicone to prevent water ingress for the general timespan a wetsuit is in the water. The structure of the sensor is pictured in Figure 2. Further details on developing a strain sensor for operation underwater can be found in a previous study.[9]

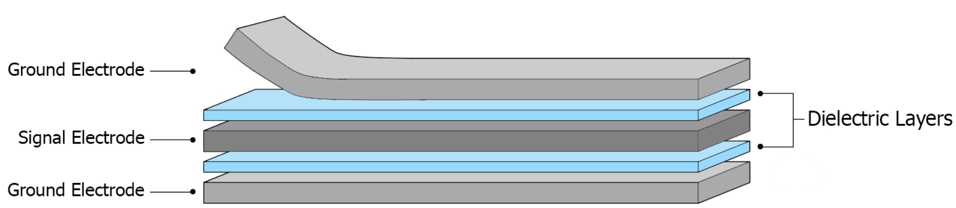

Figure 2. DE sensor structure.

The overall thickness of the sensor was approximately 1 mm allowing it to be integrated unobtrusively into underwater garments. In this study strain was defined as

$$strain = \frac{l - l_0}{l_0} * 100[\%], \tag{1}$$

where l is the current length and l_0 is the original length. The sensor had an active strain area of 6 cm (Figure 3), a relaxed capacitance of approximately 170 pF and a sensitivity of 1.7 pF/mm.

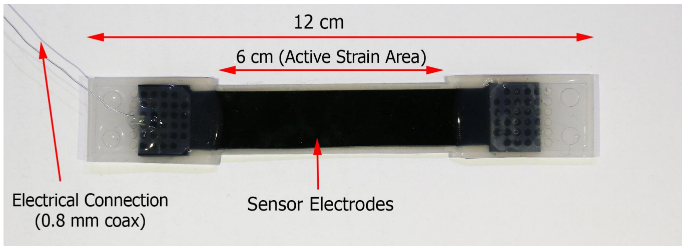

Figure 3. A DE sensor used in this investigation.

A linear motor was used to perform strain measurements in freshwater; capacitance was measured in parallel. Sensor sampling for all experiments was performed at 25 Hz. Sensors were given 20 minutes to equilibrate to the water's temperature based on results from a previous study.[9] The strain rate used was 50 mm/s with a waiting time at maximal strain of 10 seconds. The strain applied was 25%, which equates to 15 mm.

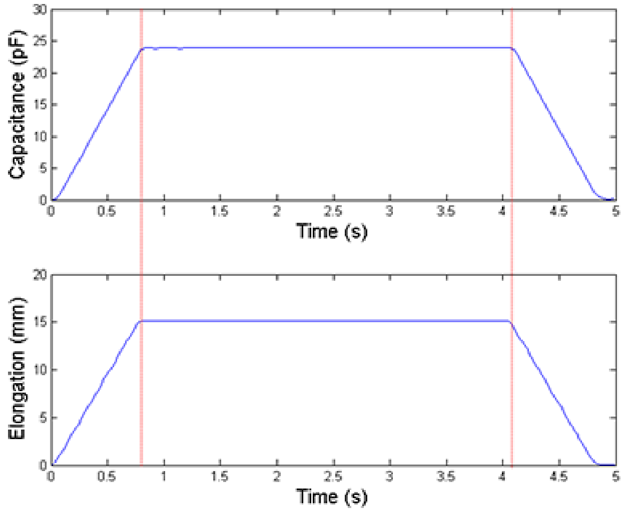

Figure 4. DE sensor characterisation showing elongation and capacitance over time.

Figure 4 shows capacitance and elongation over time for a strain sensor used in this investigation. The plot shows that the sensors do not experience overshoot or relaxation time.

The capacitance of the sensor was given by

$$C = \epsilon_0 \epsilon_r \frac{A}{t}(N-1), \tag{2}$$

Where C is the total capacitance of the sensor, ϵ_0 is vacuum permittivity, ϵ_r is the relative permittivity of the dielectric material, A is the average active electrode area, t is the average thickness of the dielectric layers, and N in the number of electrode layers.

3. EXPERIMENT PROCEDURE

Experiments took place with the participant wearing the wetsuit in a freshwater swimming pool. There were two main experiments: determining a kick count at various rates, and measuring leg joint angles. A GoPro Hero 4 silver edition camera was used to provide footage for post-analysis validation. Ethics approval was obtained from the University of Auckland Human Participants Ethics Committee (UAHPEC).

3.1 Joint Angles

The joint angles required to monitor leg motion are shown in Figure 5. α is located between the vertical plane at the hip joint and femur of the participant. β is located at the knee joint and is determined by the positioning of the fibula and femur. Orange markers were placed on the ankle, knee and hip joints to help derive angles in post-processing. With the distance between the orange markers known, the relative leg position can be calculated using α and β.

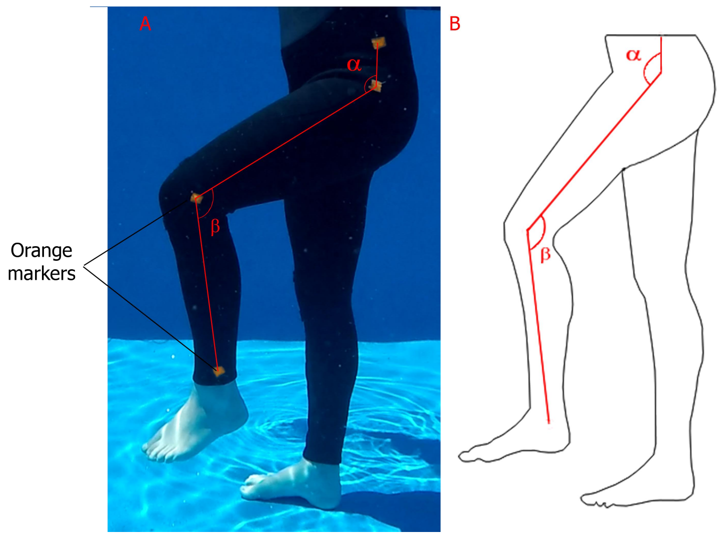

Figure 5. Image A and B show the angles of interest, α and β.

3.2 Sensor Calibration

To convert linear strain into an angle the sensors were calibrated in a two step process. The participant held their legs at two positions with known joint angles: standing straight, 180° angle for both α and β, and leg raised with α and β both forming right angles. Using the values recorded at both positions a ratio between capacitance and degrees was determined. The ratio was used in real-time to convert capacitance data into joint angles. This conversion assumed linear strain as the participant rotated their joint.

3.3 Kick Count

For this experiment a kick was defined as a relative threshold between the two legs. The participant performed kicking motion by raising one leg at a time transitioning both α and β between 90° and 180°. As one leg raised, both the α and β angles of that leg decreased. When the angle was to at least 10% below to opposite leg, a kick was counted.

Three different rates were tested during the experiment: one kick every one, two and three seconds. Each rate was performed for a period of 1 minute. The resulting kick count was validated by video analysis. The results can be seen in Section 4.1.

3.4 Leg Position

In this experiment the participant transitioned their left leg to 20 different positions. The α and β angles were calculated in real-time by connecting the strain sensors to a nearby computer. The 20 positions performed by the participant are shown in Figure 6.

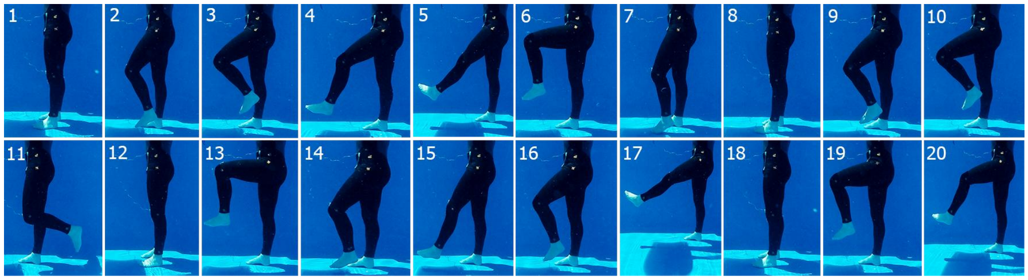

Figure 6. The 20 positions chosen at random by the participant.

The experiment was conducted with the participant holding each position for three seconds. This gave time to average several measurements both by the strain sensors and in image processing. Post-analysis was performed using ImageJ.

4. RESULTS

4.1 Kick Count

In this experiment the participant was instructed to raise one leg at a time to the position shown in image 5 of Figure 6. The sensor signal should therefore show the angle transitioning between 90° and 180°. Figure 7 shows raw data from the left knee strain sensor (β angle) for one minute periods at each rate. The dotted lines indicate the 90° and 180° levels.

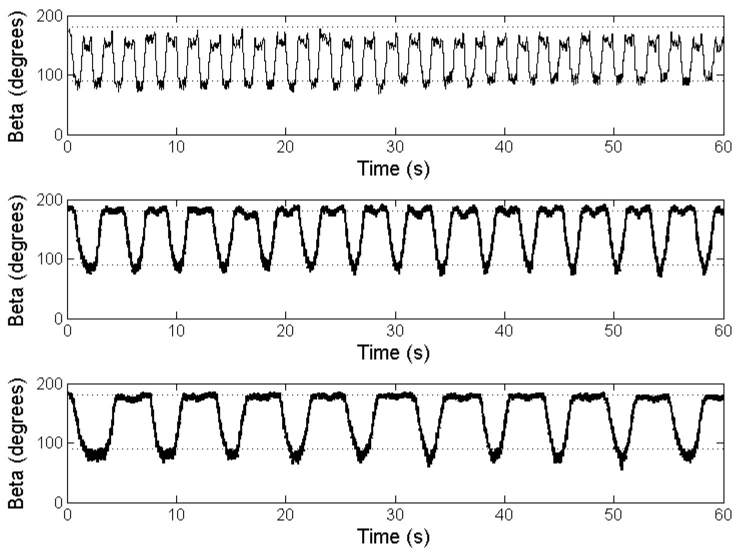

Figure 7. β angle from kick count experiment: top plot shows one kick per second; middle plot shows one kick every two seconds; lower plot shows one kick every three second.

Figure 7 indicates that the participant bent their leg further than 90° at all three rates. This is shown by the signal dropping below the bottom dotted line in all plots. Furthermore when the participant was performing one kick per second (top plot) they did not return their leg back to straight position (180°). This is visible by the signal in the top plot never reaching the 180° dotted line. The strain sensors in this experiment were able to correctly determine the kick count at all three rates despite these faults. A kick count comparison between video analysis and the strain sensor algorithm is shown in Table 4.1.

Table 1. Table comparing accuracy of kick count calculation.

	1 kick/s	2 kicks/s	3 kicks/s
Strain sensing algorithm	60	30	20
Video analysis	60	30	20

4.2 Leg Motion

The participant transitioned their left leg to 20 different positions. A comparison of α and β angles measured by strain sensors and those derived from image processing can be seen in Figure 8.

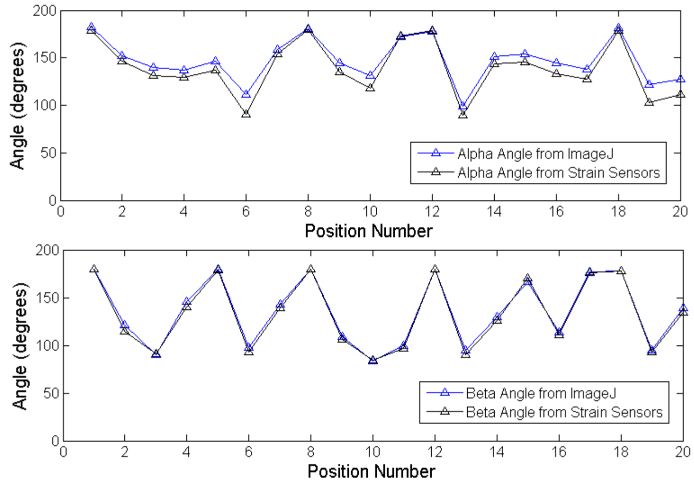

Figure 8. Comparison of the angles calculated at each position using strain sensor signals and in post-image processing.

In this experiment the knee sensor, β, performed more accurately and reliably than the hip sensor, α. The sensor signal underestimated the hip angle when the leg was raised. This is seen in the top plot of Figure 8 by the growing separation between the two lines. With respect to the knee joint however, the angle calculated from the sensor signal was, on average, within 2.2% of the angle determined in image processing. This is compared with an average error of 6.5% for the hip joint. Figure 9 plots the percentage difference between strain sensor and post-analysis across the angle range.

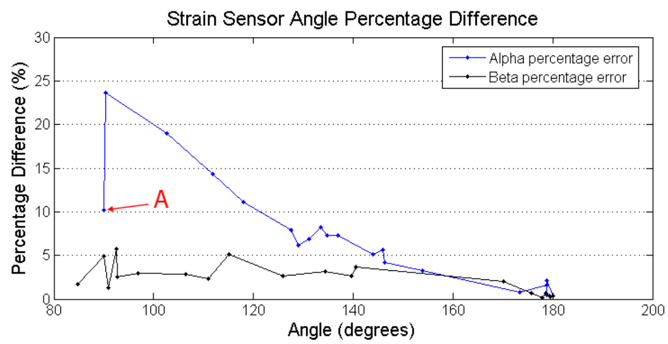

Figure 9. Percentage difference between strain sensor and image analysis angles.

With the exception of point A, there is an inverse relationship between the α angle and percentage difference. In contrast the β angle shows no significant dependence on angle. The maximum difference for the α angle was 23% and occurred at an angle of 92°. This is compared with the maximum difference for β, 5.4%.

5. DISCUSSION

The error observed in Figures 8 and 9 is not specific to this investigation, but represents wider challenges of using strain sensors to measure human motion. The internal framework of the body is a rigid skeleton which is contrasted on the surface by soft and elastic skin. Sensing human motion refers to measuring movement of the skeleton. In this investigation it was achieved indirectly by integrating strain sensors into a wetsuit worn by the participant. This is an indirect form of measurement as there is in fact three general interfaces separating the sensor from the bone: bone-skin, skin-wetsuit, wetsuit-sensor (Figure 10). Each interface is inherently a source of error.

Figure 10. As the joint rotates, the skin deforms which in turn interfaces with the clothing and sensor.

Any relative shifting between interfaces can interfere with the quality of the signal. This is seen in a study by Tairych et al. where a sensing glove began to shift position during fast gesture motions.[10] With regards to this investigation interface shift is likely responsible for the underestimation of α seen in the top plot of Figure 8. As the participant raised their leg there was a build-up of wetsuit material between the thigh and pelvis. Therefore the wetsuit shifted position relative to the skin. Furthermore as the hip sensors attached via their ends, as the participant's leg raised the active area of the sensor separated from the wetsuit. This is illustrated in Figure 11.

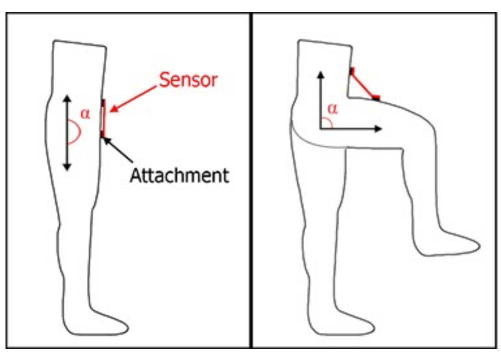

Figure 11. Positioning of hip sensor in both positions performs by the participant.

In the ideal situation as the joint rotates and skin deforms accordingly, the wetsuit and sensor should follow this deformation without hindering it. To achieve this a better understanding of interface shift and is required.

The first interface is between the epidermis, which is the outermost layer of skin, and the skeleton of the body. While the bone is relatively rigid, skin is designed to deform and stretch on command. The epidermis across the knee joint, for example, stretches between 35% and 45% depending on the individual.[11] Furthermore skin exhibits both viscous and elastic properties which also need to be taken into account when designing a strain sensing garment. The Young's modulus of skin varies across the body, between people, and with different hydration levels, however between 4.5 kPa and 8 kPa has been approximated by several groups.[11–14] Due to the nature of the first interface, minimising shift is not possible. Therefore calibration must be used to minimise the effect this has on the quality of the signal. Assuming skin deforms in a repeatable manner, calibration should be able to minimise the impact of this interface substantially. To minimise relative shift in the second interface however, the garment should have similar characteristics to skin.

If the garment has lower strain capabilities or is stiffer for example, then not only will it cause skin to deform in a different manner but the garment may also physically shift position relative to the skin. If a shift occurs then error is now introduced as the garment and therefore sensors were in a different position during the calibration procedure.

The last interface exists between the sensors and the wetsuit. Techniques should be implemented to ensure the sensors do not shift relative to the garment. O'Brien et al. reduced error by customising the sensors for a particular placement.[15] This was achieved with the development of the "finger net" system (Figure 12). The length and width of the sensors were customised to fit neatly on the wrist while maintaining sufficient width to grip the knuckles. Additionally fabric rings were added to hold the sensors in place during deformation.

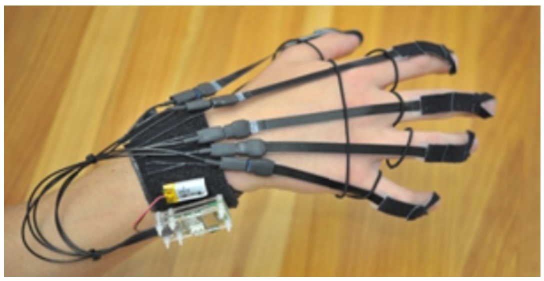

Figure 12. Stretchsense "finger net" system with integrated stretch sensors.[15]

In this study sensors were attached only at the ends. Therefore the active area was free to shift introducing error into the calibration. This attachment method was chosen as it was appropriate for prototyping allowing easy sensor replacement. The downside is it relies on the participant rotating joints in a repeatable manner. To improve repeatability sensors could be glued along their entire length or fabricated directly onto the wetsuit. This change however method relies on the wetsuit accurately following the deformation of the skin. Therefore the wetsuit would need to be customised to reduce build-up of material at the hip joint as the leg raises.

Additionally the location of the sensor could also be reviewed. The knee sensor performed well across the entire angle range despite only being attached at its ends. This is seen in Figure 8 with an average percentage difference of 2.2%. The primary difference between the hip and knee sensor was placement. The knee sensor was always strained above 180°. Therefore the sensor's active area remained stretched over the wetsuit for the duration of the experiment. This is seen in Figure 13 with the sensor staying close to the wetsuit in both positions.

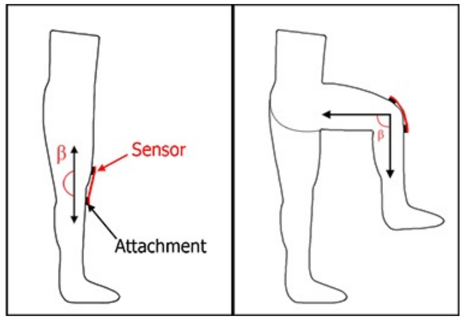

Figure 13. Positioning of knee sensor in both positions performs by the participant.

To improve hip sensor performance the sensor could be moved to an alternative side of the suit. Mengüç et al. placed the sensor on the rear side in their investigation of lower limb biomechanics.[16] Due to the nature of strain sensing a sensor on the rear side would need to attach from the lower back to upper hamstring. To achieve this Mengüç's group attached inextensible straps to their standard sensor to increase the overall length without having to custom fabricate a sensor for this placement. For a diving wetsuit placing a sensor on the left and right side of the hip joint may reduce potential for damage. These potential improvements are outlined in Figure 14.

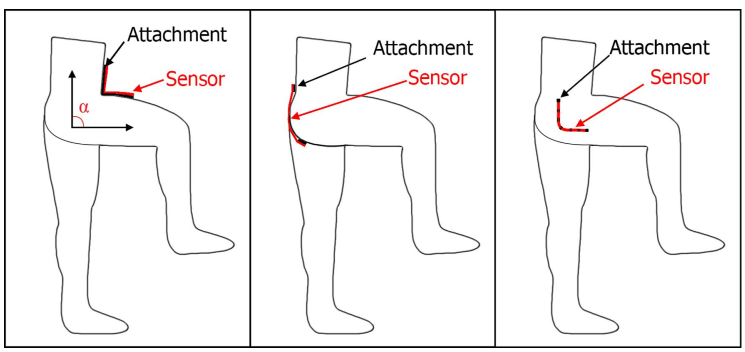

Figure 14. Improvements for measuring the hip joint.

The improvements suggested so far are focussed on minimising the relative shift between interfaces. Lu et al. proposed an alternative solution by removing one of the interfaces entirely; they mounted the strain gauge directly onto the skin.[17] For many applications however, mounting the sensor directly onto the skin may be considered too invasive or time consuming. Park et al. took a similar approach to Lu's group by removing the garment-sensor interface.[18] Park's group developed a sensing thread that could be directly integrated as part of the garment. This is ideal as it reduces the number of interfaces but does not make the measurement more invasive for the user. The garment would still have to be customised to the wearer and match the elasticity of skin for accurate results.

Despite the error present in the hip joint measurements the sensor signals were used to correctly determine the kick count of the participant. This is seen in Table 4.1 with the correct kick count identified at each rate. This implies that exact strain measurements may not be necessary for gesture recognition 90° joint rotation. This is also justified by Figure 8 where despite the underestimation in the hip joint both lines follow the same pattern. Therefore depending on the application relative shift minimisation may not be critical to the performance of the sensing garment. This experiment has highlighted that sensor accuracy may not be the main source of error in human motion measurements. While sensors perform accurately and reliably on a linear motor it is critical to consider how to minimise relative shift between interfaces right through from the design stage to implementation.

6. CONCLUSION AND FUTURE WORK

In this paper a wetsuit prototype was used to measure leg motion. We used strain sensors that had a linear relationship between capacitance and strain. 3M lock fasteners were used to attach these sensors to the knee and hip regions of a wetsuit.

The overall goal was to assess DE sensors' performance on diver kinematic monitoring. The experiment showed that sensors were able to accurately monitor rotation of the knee joint and large angles of the hip joint. Below 150^o the sensors underestimated the angle of the hip joint by up to 23%. Accuracy is likely to improve with a custom-fit wetsuit, optimised placement of sensors, increased prestrain, and an improved calibration procedure. Optical imaging could be used to find high deformation placements for each sensor position. Additionally the geometry of the sensor could be optimised for each location.

Future work includes expanding the amount of sensors on the suit to monitor arm motion. Additionally attaching strain sensors to the chest and abdomen would allow for respiratory monitoring.

This investigation has highlighted that it is not sufficient to have accurate sensors. It is essential couple sensors in a way that minimises interface shift and maintains the quality of the signal signal.

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