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Development of a Wearable Sensor System for Dynamically Mapping the Behavior of an Energy Storing and Returning Prosthetic Foot

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It has been recognized that that the design and prescription of Energy Storing and Returning prosthetic running feet are not well understood and that further information on their performance would be beneficial to increase this understanding. Dynamic analysis of an amputee wearing a prosthetic foot is typically performed using reflective markers and motion-capture systems. High-speed cameras and force plates are used to collect data of a few strides. This requires specialized and expensive equipment in an unrepresentative environment within a large area. Inertial Measurement Units are also capable of being used as wearable sensors but suffer from drift issues. This paper presents the development of a wearable sensing system that records the action of an Energy Storing and Returning prosthetic running foot (sagittal plane displacement and ground contact position) which could have research and/or clinical applications. This is achieved using five standalone pieces of apparatus including foot-mounted pressure sensors and a rotary vario-resistive displacement transducer. It is demonstrated, through the collection of profiles for both foot deflection and ground contact point over the duration of a stride, that the system can be attached to an amputee's prosthesis and used in a non-laboratory environment. It was found from the system that the prosthetic ground contact point, for the amputee tested, progresses along the effective metatarsal portion of the prosthetic foot towards the distal end of the prosthesis over the duration of the stride. Further investigation of the effective stiffness changes of the foot due to the progression of the contact point is warranted.

Keywords: Assistive devices, biomedical measurements, displacement measurement, force measurement, gait analysis, prosthetic limbs, wearable sensors.

1. Introduction

Since the commercial introduction of the Energy Storing and Returning (ESR) prosthetic foot in 1985 [1] prosthetists and amputees have been able to choose ever-more specific and potentially suitable feet for any given application. These feet are termed energy storing and returning as they store energy when the amputee lands and return this potential energy when toe-off occurs. Styles of feet have been developed that, on a competitive running stage, significantly outperform those designed for simply restoring basic function (such as the SACH foot (Fig.1.a)) or a medium-level activity foot (Fig.1.b)). Running-specific models such as the Ossur Flex Run (Fig.1.c)) have allowed performance approaching that of the highest level able-bodied athletes. However, according to Strike & Hillery [2] the design of prosthetic feet appears to have been carried out on a trial and error basis. The dynamic characteristics of these feet have been investigated [3]-[9] but often the findings raise more questions than have been answered.

a) b) c)

Fig.1. Typical a) SACH (Solid Ankle Cushioned Heel) foot for low activity (Chas A Blatchford & Sons Ltd), b) ESR medium level activity foot (Chas A Blatchford & Sons Ltd Elite II) and c) ESR running-specific foot (Ossur Flex Run).

Hafner [8] and Wilson et al [10] suggested that there is a disconnect between scientific evidence and clinical decision making. This is supported more recently by De Luigi and Cooper [11] who have noted that prescribing prosthetic components that facilitate higher activities is typically based on the experience of the prescribing physician and of the prosthetist. While Raschke et al [12] indicate the data to link mechanical characteristics to appropriate functional level or

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to user preference is incomplete, presenting a hindrance to evidence-based practice in the field. Fundamentally the dynamic nature of prosthetic running feet is not fully understood.

Hafner et al [13] conclude their studies of the clinical prescription methods by suggesting that if further work were to be conducted, it should be carried out in 'real-world' environments including stairs, hills and uneven terrain to better serve the clinical prescription of prosthetic feet. Therefore, if progress is to be made in understanding the performance and prescription of running prosthetic devices, accurate data must be collected in vivo.

If a system were developed that could unobtrusively be worn by the amputee and were able to record the activity of the prosthetic foot; this would allow data acquisition during the regular daily regime of the individual meeting the recommendations for further work from other research studies.

Whilst there are examples of wearable sensors for gait analysis [14], [15], typically dynamic analysis of gait and deflection is performed using reflective markers and two- or three-dimensional motion capture systems with multiple high-speed cameras [2]-[4], [13], [16], [17]. This approach requires highly reflective markers to be attached to specific points of the individual under examination and their motion recorded using high-speed cameras under direct artificial light. The resulting video capture can then be analyzed using specific software packages to describe, calibrate and quantify the relative movement of the marked positions. Force plates can also be used to collect data of a small number of strides [3], [4], [18]-[20]. Such an approach requires highly specialized and expensive equipment in a controlled and unrepresentative environment, within a large area for the setup of the apparatus. It has been reported by Windolf et al. [21] that there is significant influence of the system environment (camera setup, calibration, marker size lens filter) on the performance of video-based motion capturing systems.

Inertial measurement units (IMUs), which measure force and angular rates using accelerometers and gyroscopes, have the potential to been used to capture human motion in natural environments. Cheng et al [22] states, IMUs could be used for human motion capture with great portability and flexibility. They would work almost everywhere, but are unable to maintain long-term accuracy because of sensor drifting and the interference from local magnetic fields. The issue of drift is also recognized by Rebula et al [23] who have reported on the variability of the measurement of foot placement with inertial sensors due to drift.

Therefore, a new form of sensor system is required to overcome the limitations of typical gait analysis systems.

The device should:

- measure and record the required variables (to be defined) without drift.
- be small and light enough to avoid influencing the running style of the amputee.
- be able to log data for the entire duration of exercise.

This paper reports on the research undertaken to develop a sensor system to meet the above specified design criteria and makes recommendations for its future development. The system has been designed primarily as a research tool however it is speculated that such a system could be employed in a clinical environment to inform prescription of ESR feet.

2. Material and methods

A. What variables should be measured?

Following a review of the literature regarding the dynamic function of prosthetic feet, a list of desired variables to be measured was compiled (Table 1.). Prosthetic ESR feet are generally regarded as functioning primarily in the sagittal plane (a longitudinal plane that divides the human body into right and left sections) meaning that the measurement of in-plane bending is adequate for understanding their dynamics. If this primary mode of operation can be observed accurately then a number of factors can be recorded such as stride cadence, swing timing and rate of energy absorption/return.

One phenomenon that is mentioned in literature [24] but not explored is that of boundary conditions. If a single stride is examined, following foot strike, the tibia (in the case of an intact limb) progresses over the foot (tibial progression). This motion naturally transfers the weight of the runner from the extremity of the heel, forward to the toe, until the foot leaves the ground (toe-off). The same principle can be observed in an ESR running foot (for example with an Ossur Flex Foot). In this instance the lack of a heel means the initial ground contact point is on the anterior (front) portion of the foot.

As tibial progression occurs it can be observed that the ground contact point alters through to toe-off relative to the toe of the foot (Fig.2.a)). Given that the shank of the foot remains attached to the prosthetic socket at all times but the distal (ground) contact point changes, the spring rate of the foot (force and displacement relationship) must change as the length of the effective lever arm changes. The further rearwards (posterior) the contact point with the ground, the higher the spring rate must be (Fig.2.b)). Conversely the spring rate decreases as the ground contact point progresses forward onto the toe. The spring rate is a measure of the foot stiffness, the higher the spring rate the stiffer the foot.

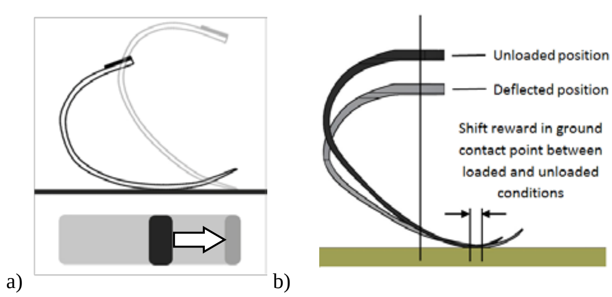


Fig.2. a) Projected ground contact point progression throughout a single stride on an ESR running foot and b) reward ground contact position shift due to ESR foot deformation.

This suggests that the spring rate of a foot is dependent on the running style of an individual. Even assuming straight level running, if the runner takes shorter strides or has a more digitgrade ('on the toes') characteristic set by the prosthetist, the ground contact will be different to that of a more relaxed user and thus a different spring rate variation across a stride will result. Contrasting styles are also evident if the same runner takes part in sprinting or jogging activities (Fig.3.). Thus the system was designed to also investigate ground contact position.

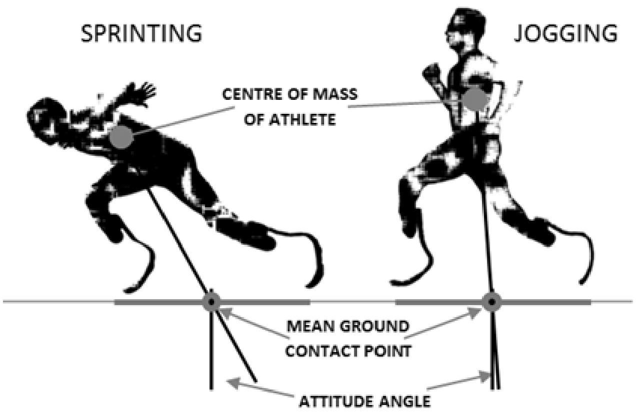


Fig.3. Different running styles will influence the attitude of the runner and hence the ground contact point of the prosthetic device, affecting the spring stiffness.

B. Apparatus

Once a complete list of variables to be measured was compiled, a range of devices were selected to allow the respective data collection. A full list can be seen in Table 1. and details of each piece of equipment in Table 2. with a brief description of each following.

Table 1. Summary of variables to be measured and specific device proposed.

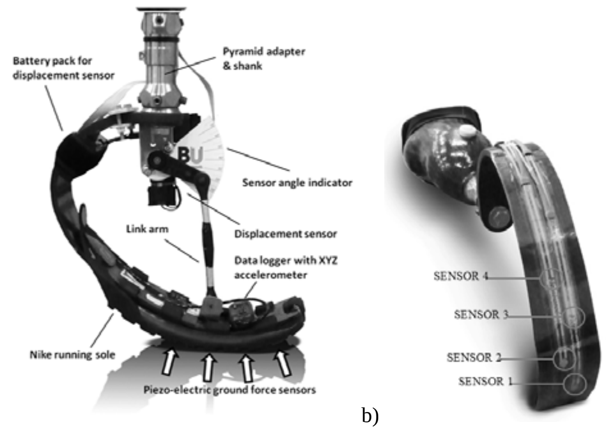
Variable	Units	Measurement Device
Stride cadence	Hz	Displacement Sensor
Ground contact time	second	Displacement Sensor
Swing phase time	seconds	Displacement sensor
Timing of maximum displacement	Seconds (after heelstrike)	Displacement sensor
Amplitude of maximum displacement	mm	Displacement sensor
Rate of energy storage	mm/s	Displacement sensor
Rate of energy return	mm/s	Displacement sensor
Ground contact point	mm (posterior of toe face)	Piezo-resistive pressure sensor array
Tri-axial acceleration of toe region	m/s^2	Tri-axial accelerometer

Table 2. Summary of equipment used, manufacturer and details of the device.

Description	Manufacturer	Details
1. Displacement Sensor	Hartmann Automotive	Hall-effect rotary sensor
2. Ground-Force Sensors	Tekscan, Inc	Piezo-resistive sensor (0-100 lb) (Flexiforce)
3. Resistive Force Signal Conditioner	Bournemouth University	4 channel signal conditioner/amplifier (0-5 V output)
4. Analogue Datalogger/accelerometer	MSR Electronics GmbH	4 channel 1 kHz 0-5 V logger
5. Battery Pack	Unknown	3 x AA alkaline cells (4.5 V)

1. Displacement sensor

As the primary mode of operation of an ESR prosthetic running foot is bending in the sagittal plane accurate logging of this mode is fundamental to understanding the dynamics of the system.



a) b)

Fig.4. Prosthetic running foot with a) sagittal plane displacement instrumentation attached and b) Piezo-resistive sensor array on metatarsal region.

For this reason, an automotive suspension position sensor (height sensor) was utilized. This is a resistive hall-effect device with a rotary input that emits a variable voltage output depending on arm position and supply voltage. Originally the purpose of such a sensor is for detecting suspension displacement and quality of road surface for the control of variable damping on the suspension systems of luxury automobiles. It has a specified manufacturing tolerance (across devices) of 2 %, i.e., a position measuring range with an uncertainty of 2 %.

The displacement sensor was attached to the proximal end of the prosthetic foot via a fabricated aluminum bracket. The bracket was lightly clamped at the carbon fiber section of the foot leaving it adjustable, easily detachable and non-invasive; as such it can be moved from one prosthetic foot to another without compromising the feet. The standard link arm of the sensor is specified for fitment to the original vehicle suspension system and as such was modified. It was lengthened in order to attach to the distal end of the prosthetic foot via an adjustable threaded turnbuckle-style system allowing accurate tuning of the length (and therefore electrical output reading from the sensor) for different prosthetic feet and set up conditions. As such the displacement sensor was able to measure any change in distance between the proximal and distal ends of the prosthetic device (and therefore a measure of strain energy). The setup of the displacement sensor can be seen in Fig.4.a.

2. Ground force sensor

In order to build up a more accurate image of the dynamics of a prosthetic running device it was necessary to understand the ground contact point throughout a single stride. If the displacement of the foot is known, then the force going through the foot can be easily derived as long as it is known where that force is being applied (i.e. what the ground contact point is).

The sensors chosen for this task were piezo-resistive devices of a printed construction from Tekscan, Inc. These units are flexible and <0.2 mm thick and therefore can be inserted between the carbon fiber foot and the foam of the trainer sole used, thus protecting the sensors from direct contact with the ground. The sensing area is only 10 mm diameter and cannot differentiate between pressures applied at various points within this sensing area. Therefore a linear array of these sensors was used (four in total) along the metatarsal region (ball) of the prosthetic foot (Fig.4.b)) and their results interpolated. A variety of these sensors are available from the manufacturer. The exact variant chosen for this investigation were the 100lb (45.4 kg) 9" (228.6 mm) model.

3. Resistive force signal conditioner

This piece of apparatus was designed and built in-house for the specific application of converting the changing resistive function of the pressure sensors to an analogue voltage signal that could then be logged. The total size of this device was 140 mm x 80 mm x 30 mm, weighing approximately 220 grams and throughout the course of the investigation it was stored in the pocket of the participant.

4. Analogue data logger

In order to capture the data being generated by the respective sensing devices, a four channel analogue 0-5 V datalogger was used. The actual device chosen was a standalone 'MSR165' model from MSR Electronics GmbH (Modular Signal Recorders) of Switzerland. It is capable of logging 4 analogue channels simultaneously at a selected frequency up to 1024 Hz but is small and lightweight enough to be placed on the foot itself and not be noticed by the amputee (39 mm x 23 mm x 52 mm, 686.7 m/s^2). The logger was configured to start and stop data acquisition with the push of a button on the outer surface of the device. As an additional function the logger also contained a tri-axial accelerometer capable of recording +/-147.15 m/s^2 to an accuracy of +/-1.4715 m/s^2 at a frequency of 1600 Hz. The logger contains its own battery and can log for many hours at maximum frequency without running out of capacity. Once the data acquisition is complete the logger can be attached to a PC via USB interface and data viewed in .csv format.

5. Battery pack

The data logger required an analogue input of 0-5 V for each of the four channels. As the Displacement Sensor was not internally powered, an additional battery pack was required. A common 3-cell AA alkaline battery case was chosen to provide an output voltage of up to 4.5 V from the displacement sensor (subject to input arm angle). To ensure repeatability the state of charge of this pack was checked at the start of each test by measuring the output voltage.

C. Equipment setup

The equipment was set up in a manner that could easily be repeated should further testing be required. A foot with all of the instrumentation attached can be seen in Fig.4. A running sole was attached to the foot over the ground pressure sensors. The sole provides stability and traction for the amputee during running. The total mass of the sensor system attached to the foot was found to be 148 grams. It must be noted that the foot used in this series of measurements had been set up by a qualified and recognized prosthetist specifically for the amputee volunteer and the adapter between the foot and shank adapter was not modified. All of the equipment was able to be fitted in a non-invasive manner so as not to affect or significantly influence the use of the foot.

1. Displacement sensor

The setup started with the proximal end of the foot (the portion that attaches to the shank adapter) being aligned parallel to the ground surface. The displacement sensor was then arranged such that the pivot of the rotary arm was directly below the centerline of the shank adapter. Once this was performed, the link arm (one end of which attaches to the distal end of the rotary arm) was attached with an adhesive pad to the toe portion of the foot such that it bridged the space between the toe and the displacement sensor. This link arm featured a turnbuckle-style thumbscrew allowing the length to be tuned for different foot geometries. The screw could therefore be used to align the radial arm of the displacement sensor with 0° on the angle indicator (an integral part of the displacement sensor bracket) and the output of the displacement sensor checked with a volt meter to confirm the sensor was at the extreme end of its range. The logger could then be wired to allow data acquisition from the displacement sensor.

2. Ground pressure sensors

Due to their fragile nature and to make their transference from one foot to another practical, the sensors were mounted on a sheet of acetate using tape to secure. They were positioned into a near-linear array with equal spacing along what was anticipated to be the dynamic contact patch of the foot with the ground during running. The front edge of this acetate sheet could then be aligned with the distal edge of the toe region of the foot for repeatability (Fig.4.b)).

Once the ground force sensors were attached to the base of the prosthetic foot they could be wired to the signal conditioner and in turn to the data logger. As such, the system could record the output of all four sensors simultaneously at a frequency of up to 1024 Hz.

It is not the intention that the sensors would provide a value of ground pressure; rather a value of contact distance from the distal end of the prosthetic device. This information was then used to illustrate the change in effective spring stiffness of the device through the course of a single stride.

D. Sensor system testing

1. Displacement transducer calibration:

To ascertain displacement (in mm) from the logged voltage, the system needed to be calibrated. This calibration was conducted using an Instron 2280 dynamic hydraulic test machine (Instron Inc.) with a 5 kN load cell fitted and a fabricated jig to mount the entire prosthetic device. The fixture was designed such that the foot was subjected to purely compressive loads, with each end clamped but possessing a single rotational degree of freedom in the sagittal plane (provided by two pairs of machined fulcrums). This meant that the proximal and distal ends of the foot had the flexibility to allow the foot to deflect naturally and not fight against the fixture as the load increased. This is demonstrated in Fig.5., shown with the instrumentation removed.

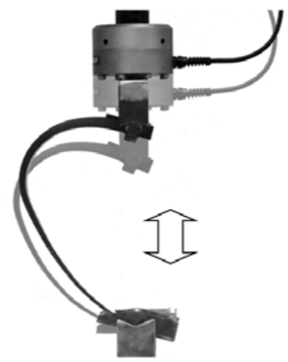


Fig.5. Representation of prosthetic foot displacement during calibration (instrumentation removed for clarity).

This approach was adopted to ensure that any load readings taken were purely as a result of the spring characteristics of the foot and not a build-up of friction somewhere in the mounting hardware.

The foot was subjected to a slow (0.2 Hz) displacement through the working amplitude of the prosthetic device (72 mm) using a triangular wave. Load and displacement data from the linear transducer of the machine was captured using the Instron DAX software at a sample rate of 100 Hz, whilst simultaneously voltage data from the rotary transducer due to its displacement was recorded using the foot-mounted instrumentation and MSR logger at a sample rate of 128 Hz. The displacement data from the machine was collated with the voltage output from the rotary transducer against time for a series of three deflection cycles (SD = 0.004 V and 0.08 mm). This was then averaged into a single deflection of the foot through the stroke of 72 mm. The calibration curve for the rotary displacement transducer was plotted as voltage output against linear deflection in mm (Fig.6.).

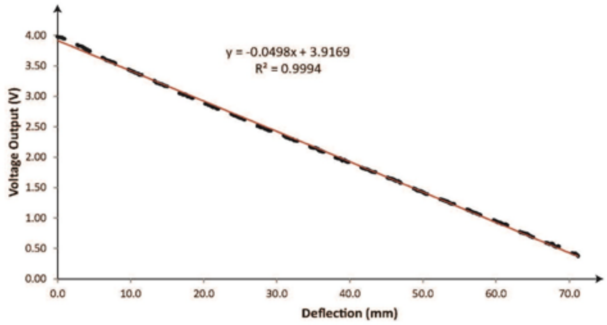


Fig.6. a) Typical calibration curve for rotary transducer (dashed line), and b) trend line (continuous line) [SD = 0.004 V and 0.08 mm].

2. System testing:

The system was tested with a long-term and regular user of an ESR prosthetic foot who did not suffer from extreme or influential pathologies such as restricted movement or chronic pain that might adversely affect running style or repeatability. The participant was a 32-year-old male left-side unilateral trans-tibial amputee who had been using a prosthesis for over ten years following a trauma. The participant had been the user of a category 6Hi Ossur Flex Run for leisure and fitness every day, had retained full joint articulation and suffered from no long-term pain or discomfort. They had a mass of 83 kg and as such used the correct stiffness category of foot according to the manufacturer's literature [25]. The selection of the participant and testing was conducted following Bournemouth University ethical approval (Reference ID: 4731).

The ESR foot used for the testing was a replica (identical model and stiffness category) of the foot already used by the amputee. Therefore he was familiar with the device and his running on the foot was comfortable. In order to ensure parity between the new model as tested and the aged historic foot as used by the runner, a static deflection test was conducted with both feet prior to the test taking place. The methodology mirrored that of the displacement transducer calibration testing (Section D1). The two feet were shown to have a static spring rate of within 1 % of each other.

Despite the near-identical nature of the substituted foot, the amputee was allowed 30 minutes to warm up with the foot in the test environment (a 25 meter sports hall with wooden floor) to ensure the additional mass of the instrumentation (148 grams) would not cause any notable issues. The testing routine consisted of the sustained running of ten lengths of the hall (250 m with nine turns) with the entire sequence logged at a frequency of 128 Hz. The participant was allowed to choose his own pace and cadence with which he felt most comfortable and familiar. Both sagittal plane displacement and ground contact force were collected during the testing.

3. RESULTS

A. *Sagittal plane deflection*

The acceleration, deceleration and turning portions of the data acquired were not considered in the analysis. A three-step portion at the center of each run was isolated and the mean values calculated resulting in data for three averaged strides (SD = 0.021 V). This data is shown in Fig.7., displayed as the output voltage from the deflection sensor versus a time trace in minutes: seconds. This value of foot deflection was not calibrated into millimeters because doing so would falsely simplify the action of the foot. The value of deflection as a function of voltage reflects the position of the lever arm of the deflection transducer and as such a trend of foot deflection as measured at the point where the link arm meets the foot keel. However, the keel deflection at any other point is different to this attachment point; the value of foot deflection as a function of millimeters depends on what point along the keel is being tested. Retaining the deflection as a function of voltage at this stage avoids misinterpretation of the results.

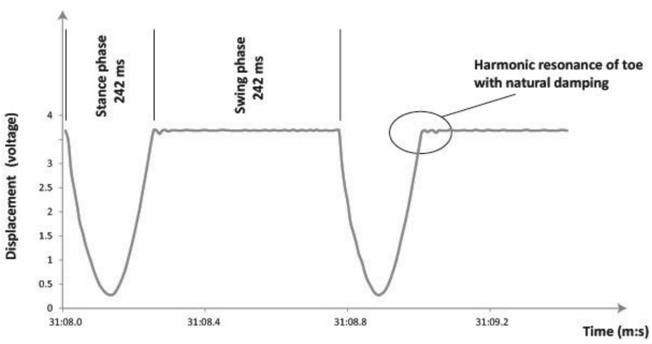


Fig.7. Trace showing the deflection characteristics of the foot tested (averaged raw data from rotary transducer with output in volts. SD = 0.021 V).

Clearly visible is the timing sequence of the runner with well-defined stance and swing phases as well as the rate of deflection and energy return. The stance phase (occurring from heel strike to toe-off) occurs over a period of 242 ms with the swing phase lasting 486 ms until heel strike for the next stride takes place. Worthy of note is the clear demonstration of the natural harmonic frequency of the unloaded foot when toe-off occurs. The trace can be seen to resonate, diminishing with the natural damping of the device (provided by losses in the system such as air resistance and friction within the foot keel).

B. *Ground contact point progression*

The data was once again averaged across all of the ten runs of the hall (ignoring the acceleration, deceleration and turning portions) resulting in a single, typical, ground contact profile. Traces for each of the ground contact sensors can be seen in Fig.8. The standard deviation of peak voltage readings for the 10 runs were calculated (SD Sensor 1 = 0.070 V, SD Sensor 2 = 0.056 V, SD Sensor 3 = 0.302 V, SD Sensor 4 = 0.021 V). In order to put the data into context, the displacement data from a similarly typical stride is overlaid to help visualize the heel strike and toe-off phases. At the time of peak pressure for each sensor, the effective ground contact point must be at this same location. Therefore, if the timing of the peak pressure for each sensor is recorded this can be plotted against its location (relative to the distal edge of the foot) to provide the trend of ground contact progression.

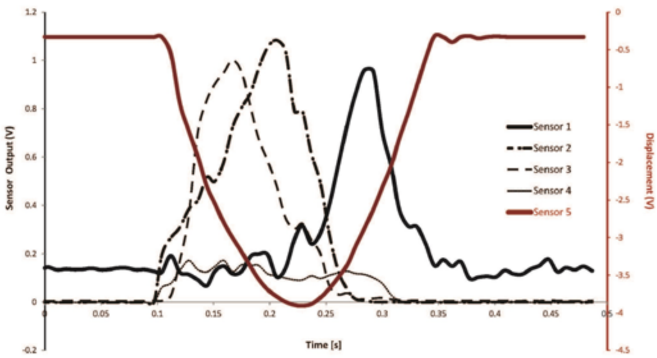


Fig.8. Force sensor outputs for a single stride with foot displacement overlaid.

It is clear from this plot (and for this particular stride with this specific foot) that only three of the sensors were useful (sensor 4 appears to represent only a baseline of contact). This is the posterior-most sensor and suggests that it is inheriting an input from the shoe sole attached to the foot but does not come into contact with the ground at any time. However, the remaining sensors show a clear progression of the peak pressure with distinct and ordered outputs. Sensor 1 also suggests an error as it has a baseline output even through the swing phase of the stride. This was investigated and is a result of its position on the toe. At this location, the profile of the foot forces the trainer sole into a curve which inherently exerts a force onto that position where the sensor is attached. Following testing the sole was removed and the sensor output naturally returned to zero. This experience identifies that careful consideration of the positioning of the sensors would be required for future experimental testing with the system.

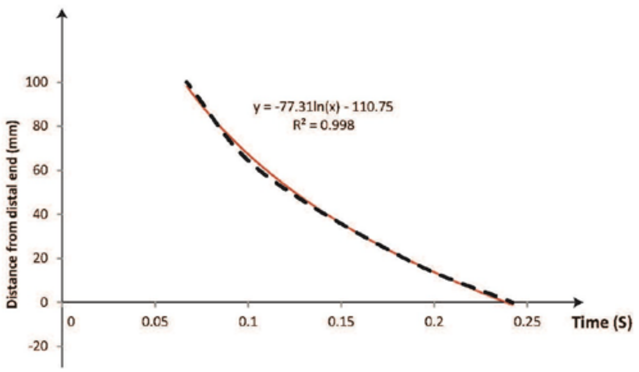


Fig.9. a) Curve demonstrating the shift in ground contact position relative to the distal end of the foot (toe region) (dashed line), b) trend line (continuous line).

From Fig.8. it is clear that the foot first contacts the ground in a position between sensors 3 and 4 (between 110 and 130 mm from the distal end of the foot) and toe-off occurs at a point forward of sensor 1 (within the end 20 mm of the distal end of the foot). Extrapolating the timing of the peak forces at the various positions it was possible to begin to build up a complete picture of the ground contact point progression and the extreme points could be found by continuing the trend line at either end (Fig.9.). It is clear that the contact point for this amputee and foot combination produced a non-linear profile. Further research is warranted to investigate the effect of ground contact progression on foot performance.

To visualize the contact progression, a high-speed video was captured of the stance phase of a single stride from heel strike to toe-off (Fig.10.). The images show the foot first contacting the ground in the region between sensors 3 and 4 (as expected) and toe-off occurs at the distal end of the foot, forward of sensor 1.

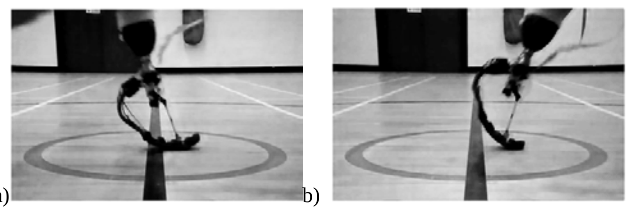

a) b)

Fig.10. Still frames extracted from high-speed video (240fps) taken of a single stance phase during data acquisition showing a) heel strike and b) toe-off.

4. Discussion

This investigation demonstrated profiles for both foot deflection and ground contact point over the duration of the stride. For the amputee on test with the specific foot used (an Ossur Flex Run Cat6.Hi) the foot response time (ground contact time) was 242 ms with a symmetrical deflection profile. Mid-stance (maximum deflection) was achieved at 141 ms. The ground contact point can be seen to progress non-linearly along the effective metatarsal portion of the prosthetic foot towards the distal end. It should be noted that this method was not designed to capture the ground reaction force but instead provide a trend of how the contact point moves. Indeed, a more accurate picture would be difficult to achieve with this method given that the foam running sole that was attached the foot offered a significant degree of load spreading onto the carbon section. For further testing, depending on the geometry of the specific foot being tested and the nature of the runner it is conceivable that these piezo-resistive sensors could be located in alternative positions along the path of ground contact in order to provide more data points against which to plot. Further work is also needed to determine whether the sensors would stand up to repeated use without the sole being attached to the foot to negate its effects on the results obtained. If accurate ground reaction force were required, calibration of the sensors would need to be carried out. This could be achieved using the Instron 2280 dynamic hydraulic test machine and a sliding plane (with adjustable height) for example.

The instrumentation that was installed on the foot as it exists in Fig.4. allowed all of the desired variables to be recorded whilst also allowing the amputee athlete as much flexibility as they would ordinarily have had. Grabowski et al [26] notes that Some Paralympic sprinters regularly add 100-300 grams over the forefoot region of their foot during competition because they feel it helps them achieve a more symmetrical gait. However, the participant was unable to notice the change in mass (additional 148 grams) in this study. Future study of the effect off adding mass to the foot is warranted.

This system allows the freedom to collect data outside of a gait laboratory that previously was not possible. Additionally, because of the high capacity memory of the MSR datalogger, many hours of data acquisition are achievable. Although a sample rate of 128 Hz was chosen for all the investigations conducted on this occasion, the logger also retains the potential of increasing this to over 1 kHz should any additional resolution be required. Furthermore, the accuracy of such a system is easily defined and repeatable due to the mechanical nature of the components involved. There is no concern of reflective markers shifting, as with traditional gait analysis, and an almost indefinite number of consecutive steps can potentially be recorded over a multitude of surfaces and gradients.

The apparatus developed in this study was able to repeatedly map the deflection of the prosthetic foot on test, but was not without its drawbacks.

- The non-permanent connection method of the mechanism to the foot can be unreliable and the logger is somewhat vulnerable in its current location on the toe.
- Set up of the apparatus can be a time consuming and delicate process. Specifically, the tuning of the displacement sensor (ensuring the output is at zero by adjusting the length of the link arm) and requires a volt-meter.

Future work would involve increasing the number of pressure sensors to allow a more accurate representation of the ground contact progression. Now that a picture of ground contact progression has been developed, the next logical exercise is to understand how the static spring rate of the foot being tested varies at different effective contact points. The fixture used during the calibration phase of this investigation could be positioned at various locations along the metatarsal region of the foot to mimic the different contact points. Therefore the varying spring rate of the foot as a runner's stance phase of each stride progresses could be ascertained.

There are also many ways to conceivably develop such a sensing system to enhance functionality. If the ground force sensors were arranged laterally across the foot instead of in a linear array, for instance, this would inform the researcher and/or prosthetist of the load distribution across the foot. This could potentially enhance the initial setup of prosthetic devices and potentially enhance comfort and/or performance. The system could also be developed to display live properties of the foot, data from the previous stride and wireless transfer to a laptop or tablet device, rather than having to manually download data with a cable at each juncture. Such an approach has been adopted previously in studies with great success [15], [26].

The merits of developing a wearable sensor for recording the dynamic characteristics of a foot are undeniable and many. But it is important to appreciate that it is not just the characteristics of the foot that are being recorded with such a setup; moreover, the characteristics of the entire system including the prosthetic device and the amputee. No doubt the results of any investigation would be different if the runner were to choose to take longer or shorter strides, or change his or her running style. Equally the data gathered from the system would be different depending on what ground is being run over. The amputee might choose to run through the forest, along the beach, up steps or along a smooth flat running track. It is here that the real value of such a system lies. Understanding can be built up around real every day running activities, not just limited to a sterile laboratory environment.

5. Conclusion

The objective of this research was to develop a proof of concept for a wearable sensor system that is capable of measuring and recording a number of key variables for dynamically mapping the behavior of an ESR prosthetic foot. It was demonstrated that the developed system could collect both foot deflection and ground contact point data while an amputee was running over an extended period of time unrestricted. The system was shown to have a high level of repeatability and was light enough for the amputee not to be noticeably influenced by it while running.

Future work will focus on ensuring the system is robust enough for repeated use and fully validating the system to ensure it does not influence the running style of the amputee. The initial results obtained have highlighted that the effect on the performance of the foot due to the change in ground contact point through the duration of a stride should be further investigated.

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