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Dickinson, Matthew

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A Simulation Study into the Use of Polymer Based Materials for Core Exoskeleton Applications

Matthew Dickinson

Abstract—A core/trunk exoskeleton design has been produced that is aimed to assist the raise to stand motion. A 3D model was produced to examine the use of additive manufacturing as a core method for producing structural components for the exoskeleton presented. The two materials that were modelled for this simulation work were Polylatic acid (PLA) and polyethylene terephthalate with carbon (PET-C), and the central spinal cord of the design being Nitrile rubber. The aim of this study was to examine the use of 3D printed materials as the main skeletal structure to support the core of a human when moving raising from a resting position. The objective in this work was to identify if the 3D printable materials could be offered as an equivalent alternative to conventional more expensive materials, thus allow for greater access for production for home maintenance. A maximum load of lift force was calculated, and this was incrementally reduced to study the effects on the material. The results showed a total number of 8 simulations were run to study the core in conditions with no muscular support through to 90% of operational support. The study presents work in the form of a core/trunk exoskeleton that presents 3D printing as a possible alternative to conventional manufacturing.

Keywords—3D printing, Exo-Skeleton, PLA, PETC.

I. Introduction

THE trunk exoskeleton is a relatively new area of research within the world of exoskeleton development. The technology offers huge potential to aid in the process of rehabilitation. The trunk exoskeleton is designed to reduce spinal loading, thus offers the potential to minimise patient back pain, hence could aid in recover after spinal injury and offer spinal stability to patients with muscular ailments [1]–[6]. Back pain, and in particular power back pain, has been said to be one of the leading reasons for disabilities on a global scale [7], [8]. When addressing these cases, surgery is only ever recommended in the most severe of circumstances [9], [10]. The trunk exoskeleton has the potential to offer an improved quality of life for millions of people around the world. Even though the use of exoskeleton technology is still in the infancy of progress, the problem of accessibility still resides, at present the costs of these devices can be as high as £35,300 per system [11], offering these types of technology's to low income countries can prove to be extremely difficult. One possible way of lower the price of these skeletons is to consider the costs for manufacturing and ease of maintenance.

Since the early 1980's when Charles Hull introduced the world first 3D printer [12], 3D printing has become a major interest in the research and development environment. These machines have become widely available due to the massive reduction in price, With the development of higher performing materials, more researchers are starting to use of 3D printed materials for orthotic and prosthetic applications [13]–[15]. Zuniga presented a study that examined the use of Polylatic acid (PLA) as a material for manufacturing of open source prosthetic limbs. The study successful demonstrated the use of PLA for the manufacture of prosthesis could serve as an alternative to conventional high priced material however, the conclusions did note the problem that PLA has poor thermal performance and noted that fluctuations in heat can lead to components failure [16]. In an effort to address the low thermal performance of the material Bo-Hsin presented a solution of modifying the –NCO (Cyanate) and the –OH (Hydroxide) ratio of the material, results in glass transition temperature (T_g) improvement from the standard 55^{o}C to 64^{o}C [17]. With these advancements further development into the use of 3D printed materials within the medical world have been untaken, more specifically single limb orthotics and upper limb prosthetics [18].

Day presented work which examined the use of 3D printing technology's for providing a unique approach for assistive devices within the clinical environment. during the work the team fabricated custom prosthesis, by using computer aided design software (Autodesk Fusion 360) to produce the parts which were then verified by using finite element analysis tools found within the same software package. These where then fabricated and tested. To establish the patients satisfaction of the devices, Pugh's Matrix™ was used. A cost effectiveness study was also conducted which compared between the 3d print process used and traditional manufacturing. The results of the study confirmed that the satisfaction of the patients was high and also the cost of manufacturing of these parts was reduced by 56% [19].

This paper builds on some of works works presented, to display the use of the 3D printed material to act as a passive support structure used in a trunk exoskeleton. All 3d geometry presented in the work was produced using the computer aided design package Autodesk inventor, for all of the finite element analysis the software package Comsol multiphysics 5.6 was used.

II. Design

The initial design challenges for the exoskeleton were to develop a system that would create enough lifting force to be able to provide motion to the human trunk. To calculate this, the trunk was assumed to act as a rotating arm acting around a lower pivot point as seen in Fig. 1.

This was mathematically represented as the net force (F_p) from the pelvis which was given by:

M. Dickinson is with the School of Engineering, University of Central Lancashire, Preston, Lancashire, PR12HE, UK (e-mail: MDickinson1@uclan.ac.uk).

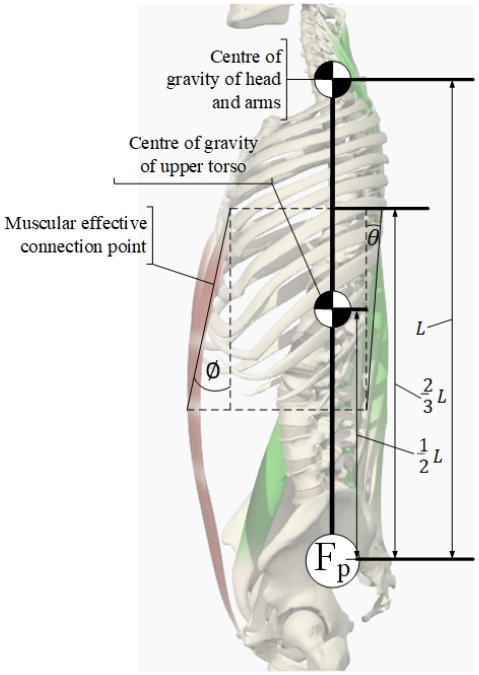

Fig. 1 Trunk Displacement

TABLE I
PLA MATERIAL DATA

Youngs Modulus (E)	3.6	GPa
Poisson's ratio (ν)	0.4	
Density (ρ)	1.27	MPa
Tensile Strength (σ_y)	49	MPa
Compressive Strength (σ_c)	18.5	MPa

$$F_p = \frac{W_h(cos\theta) + W_t((\frac{1}{2}L)\cos\theta)}{(\frac{2}{3}L)\sin(\phi)} \quad (1)$$

where W_h and W_t are the weight of the upper torso in relation to the centre of the spine and the weight of both head and arms respectively. L is the length to the head and arms centre of gravity. θ and ϕ are the bending angle and angle of the muscle respectively. The force located at the base of the spine is given by:

$$F_g = W_t \sin\phi + W_h \sin\phi + F_p \cos\theta \quad (2)$$

These forces were applied to all eight segments of the exoskeleton, this ensured a more realistic body distribution when simulated. The exoskeleton design makes use of the contour of the human trunk, each spine segment is designed in compliance to the human profile of the back, thus aids to wards comfort of wearing such a device (Fig. 2). The design of the system is broken down into three key areas which have been defined by the groupings Thoracic, Lumbar and Coccyx of the human spine. Within each area a linear actuator is placed to provide the motion to the grouping.

Each segment was produced from PLA and 3D printed with 60 % material infill. These components were connected by two 6 mm Nitrile (NBR) cables which are passed through each segment by means of two holes.

Table I shows the mechanical data used to represent the PLA in the simulation. As the design called for two linear actuators

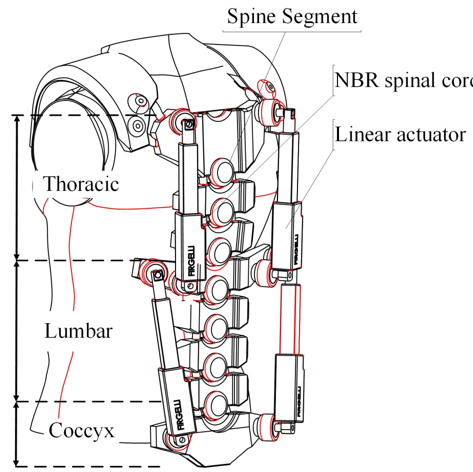

Fig. 2 Trunk Exo Skeleton

to be placed, it became apparent that if the actuator was to drive force straight into the trunk further muscular damage could occur. To combat this problem a connection solution was proposed.

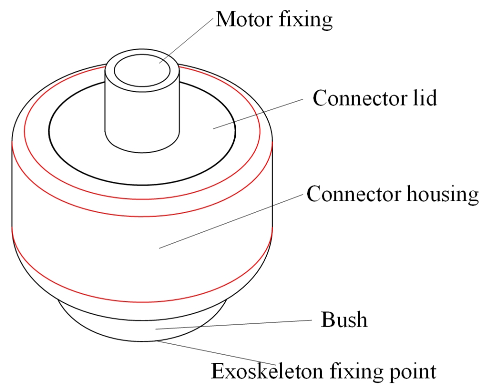

Fig. 3 Connector

Fig. 3 shows the design of the connector, the design consists of three major components, the connector lid, housing and a NBR bush. The Actuator is fixed to the lid, which is held by the housing, this assembly allows for fixing to be applied at both actuator and segment. The housing is support by bushing which creates defamation where needed. As force is applied by the actuator, a reactive trunk force occurs. As this force is not distributed equally in respects to the actuator the connector will deform thus creating an equilibrium around the holding points of the segment. Both the lid and the housing where manufactured from PLA.

As the patient moves towards a sitting position the exoskeleton is designed to move around the body creating a cradle to assist in this action. During this process the NBR cable will deform thus allowing for a more fluid motion from the body. The hyperplastic property of the cable is given by:

$$\psi(C_{10}, C_{01}, k) = C_{10}(I_1^*) + C_{01}(I_2^* - 3) + \frac{k}{2}(J-1)^2 \quad (3)$$

where I_n^* is the elastic strain coefficients with incorporation of the Helmholtz free energy per unit and k is the bulk modulus of the material. To incorporate the nearly incompressible behaviour of NBR, the Mooney-Rivlin method was used, noted in equation 3 the theorem operates with two parameters C_{10} and C_{01} which are the none linearity component and the representation of elastic behaviour of the material respectively, these have been extracted from previous work seen by de Sousa [20]. To aid the recording of the simulation the exo skeleton was defined in spinal segments from 1-8 as shown in Fig. 4.

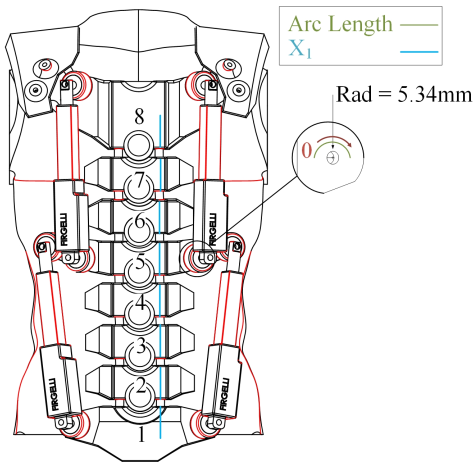

Fig. 4 Exo-Skeleton segment numbering and key identifiers

Due to the design of the skeleton being based on its ability to deform around the patients back, to examine the effects the active and passive segments on the skeleton a variation of assisting support was applied between 10-100 %, where 100 % is assuming the patient has virtually no muscle mass.

III. Results

To establish the most accurate representation of the human body creating resistance to the system while rising to a sitting position, the connector was the first component considered. The actuator places a direct force into the lid of the assembly, this is then converted to a moment of force that acts upon the body. Within the simulation the NBR was presumed to be static and the maximum capacitor of the actuator was applied, these results were transferred to the skeleton set-up.

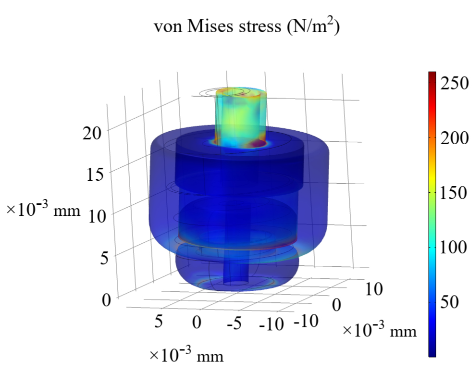

Fig. 5 Connector stress

Fig. 5 shows the results from the analysis of the connector assembly. Ranging between 250 and 50 N/m^2 the results clearly indicate that under these conditional loading's the material remains well within its recommended operating states. The results also show that the highest stress consecration are found around the base of the Motor fixing, the Connector housing and the Bush.

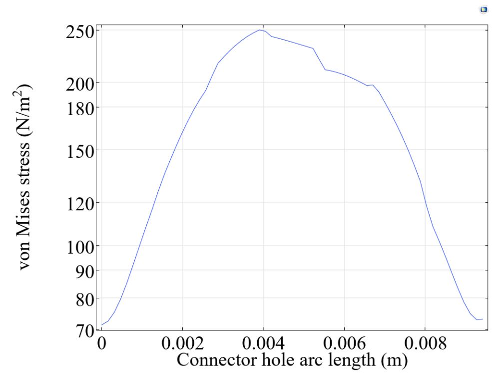

Fig. 6 Stress around lid connection

The Bush is the main connecting point between the exoskeleton and the connector thus understanding the force distribution around the connecting hole is vital to establishing an accurate simulation of the Assembly which can be seen in Fig. 6. As anticipated, the force distribution from the actuator has a smooth transition however, due to the actuator being positioned with a vector that is of centre relative to the back, the distribution of the load is uneven, where the distribution area has a smooth transition between 0 - 2 mm and due to the nature of the material infill shows distribution between 4 – 6 mm. These results were transferred from the connector positions through to the exoskeleton mount points by use of the multiphysics function with COMSOL. The results produced

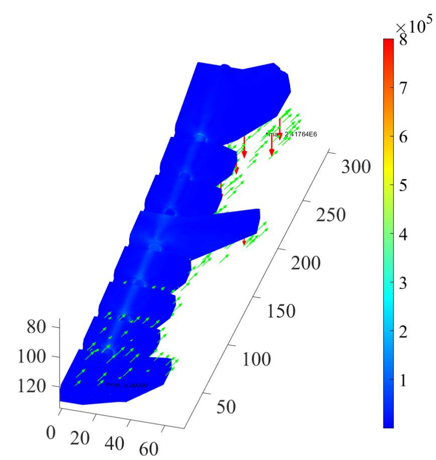

Fig. 7 Skeleton Stress (result from 100 % body mass assistance)

from the simulations of the connector are then transferred to the trunk via a means of multiphysics function within the simulation package. In this layer of the simulation force from the body is now introduced as a counter force (red arrow) to the linear actuation (green arrow), Fig. 7. The largest concentrations of stress appear around the base of the spine and also at segment 5 shown in Fig. 7. To understand the load relation across the full system the X_1 was plotted.

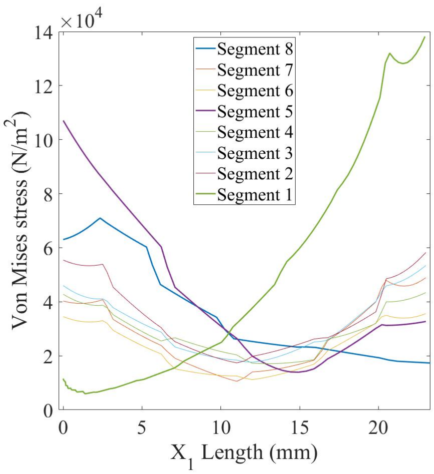

Fig. 8 X_1 variable stress plot

Fig. 8 shows the X_1 simulated stress plot, as anticipated the Lumbar and Coccyx areas both result in the highest stresses within the assembly. To deepen the understanding of the assembly and its ability to support the human body, the relationship between the Lumbar and the Thoracic was examined further, by looking at the standard deviation and also the mean of the stress. For both groups the results of the X_1 tests where examined by their mean average to offer further insight to the skeletons ability to support the body successfully.

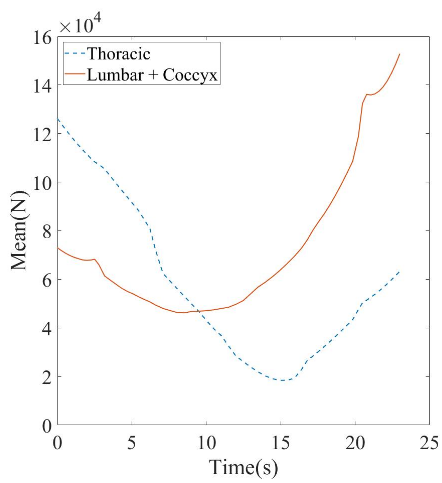

Fig. 9 Mean of both group segments

Fig. 9 shows the mean average of the two major areas of the exoskeleton the Thoracic and the Lumbar and Coccyx, the Lumbar and Coccyx show a higher distribution of mean average in relation to stress distribution, the Thoracic still do show a large fluctuation of the stress distribution much lower when compared against the top group.

Fig. 10 shows the results from the stress results of both group considering the standard deviation. The Thoracic group clearly shows a greater standard deviation that what is seen within the Lumbar and the Coccyx parts of the structure.

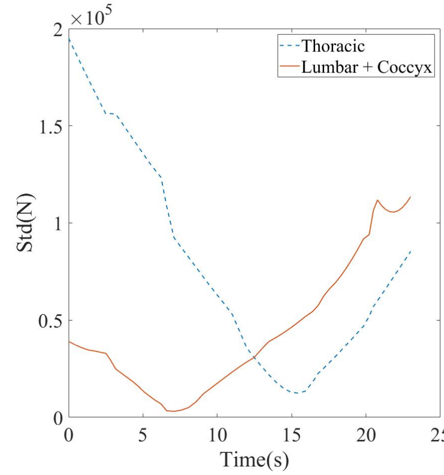

Fig. 10 Standard deviation of top and bottom groups

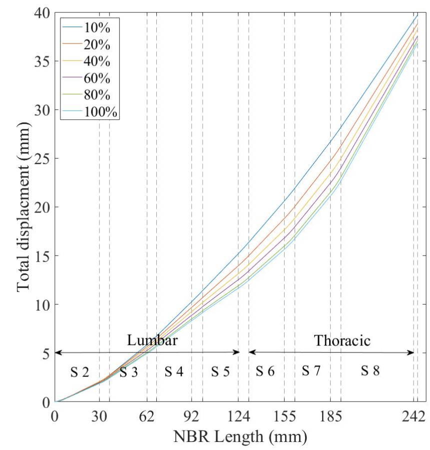

Fig. 11 Stress and Displacement of NRB cord

Fig. 11 shows the simulated results of the total displacement of the NBR based on the level of human assistance needed where 10 % would be low lever assistance needed, through to total support of 100 %. All segment locations are identified by the corresponding initials. As the Coccyx section is presumed to be the fixed position, displacement was not recorded. At 10 % assistance the response of the displacement show a smooth arc however, when the further assistance needed the transitions between the Lumbar area and the Thoracic becomes more defined, where two arcs can be seen to be forming.

TABLE II
CURVE DIFFERENCE BETWEEN 10 % TO 100 %

Assistance	20%	40%	60%	80%	100%
mm	1.97	3.27	4.25	5.20	5.60

Table II notes the maximum differences between curves seen in show in 11. The maximum and minimum deviation from assistance of 10 % ranges between 1.97 to 5.60 mm, suggested a slightly different experience of the user when committing to an action of this nature.

IV. Conclusion

This work studied the use of PLA as a material to act as geometric structure for exoskeleton components, the work also studied the use of passive and active components with the focus of supporting the complex area of the human trunk. Within this study the motion of the human body from a lying horizontal position and then moving towards a sitting state was considered. Based on the tensile strength of the material the results from the simulations showed that the material under the value of 49 MPa, hence confirming that the material has the ability to operate in the assistive technology like exoskeletons. The results showed the distribution of stress through the device as the body is rising, the Lumbar and Coccyx area shows a higher mean distribution indicating that as the body rises, the force is more focused at these group areas of the exoskeleton.

The standard deviation of the Lumbar and the Coccyx group show a much greater distribution of force throughout the assembly, suggesting that due to the drive points of the exoskeleton force is not being evenly distributed. Although the distribution of stress in the lower group is higher than the actual motion of the passive and active components showed a largest displacement of the active components was 5.60 mm when the human mass was in need of 100% assistance. These results suggest the design would serve extremely well in applications such as assistance from rising from a chair through to rising from a hospital bed.

V. Further Work

The results have shown that PLA will serve as an ideal candidate to act as the main geometric structure for trunk exoskeleton design. Due to the limiting rage of motion provided in the design, further degree of freedom should be considered. The linear actuators provided a large linear force through structure however, when applied to the connector apparent losses can be seen thus, the system will need to consider power usage, these losses would prove to have determinative effects on a mobile device such as an exoskeleton, hence this design would best be applied where power consumption would not be a problem, the examination into application of alternate actuation methods should be considered.

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Dr Matthew Dickinson is a senior lecturer in mechatronics engineering at the university of central Lancashire and an active member of the MedTech Solutions group. Originally achieving his PhD in the topic of tribology, which he was able to publish multiple conference proceedings and journal articles. Since this Dr Dickinson's work has found him changing focus from this subject and moving into smart assistive technologies, in particular exo-skeleton design. By using his skills is advanced topology Dr Dickinson aims to use accessible materials for production of low-cost maintenance of components./ Still in its early stages' Dr Dickinson has been working on use additive layer manufacturing and topological optimisation to generate high performance components that offer the same benefits as what can be seen in their traditional metallic counter parts. With his team they have also been looking at the use of machine learned for signal response, more specifically collection of electromyography muscular responses. Dr Dickinson and his team are presently partnered with the test machine company Tinius Olson who have granted access to their machine. Alongside of this work he works with ASTM and is the subcommittee chair F48.04 – exoskeleton maintenance and disposal.