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The Prediction Of The Hip Arthroplasty On Parameters Of Femur

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Abstract: The prediction of the hip joint force is a fundamental factor for the prevention of edge loading in total hip arthroplasty. Naturally, the loading of the liner of the acetabular component depends on the HIP JOINT FORCE acting on the artificial joint. In contrast to dynamic musculoskeletal models, static models for HIP JOINT FORCE prediction do not require motion analysis of the patient. However,static models have to be scrutinized for patient-specific adaptability and validity . In this study, a modular framework for HIP JOINT FORCE prediction using static models is introduced to compare the results of different cadaver templates that are the basis of most static and dynamic models, and their different scaling laws for the patient specific adaptation one-leg stance and level walking.

Using MATLAB and Three cadaver templates with varying levels of detail were integrated into the framework. A modular framework for HIP JOINT FORCE prediction using static models is introduced to compare the results of different cadaver templates that are the basis of most static and dynamic models, and different scaling laws for the patient-specific adaptation with in vivo HIP JOINT FORCE of ten patients for one-leg stance and level walking. The results revealed the significant effect of the underlying cadaver template used for the prediction of the HIP JOINT FORCE.

Using the hypothesis that a more detailed patient-specific adaptation of the osseous morphology improves the prediction significantly could only be confirmed partially for one of the cadaver templates. There are Three cadaver templates with varying levels of detail were integrated into the framework. The Fick1850 cadaver template contains 19 hip muscles represented by 44 fascicles modeled as straight lines (Fick, 1850). The Dostal1981 cadaver template contains 27 hip muscles represented by 27 fascicles modeled as straight lines (Dostal and Andrews, 1981). The Fick1850 and Dostal1981 cadaver templates were considered in this study

***Index Terms* - prediction of the hip proses, HIP JOINT FORCE patients for one-leg stance and level walking.**

I. INTRODUCTION

In this study scenario, it was implicitly assumed that a tool capable of intraoperatively measuring the soundness achieved by the implanted stem would consistently improve the success rate. This tool (software) should be accurate and during the transplant should immediately show the surgeon the suitability of the transplanted strain. Over the last 20 years, the demand for orthopedic medicine has increased as the number of patients has increased year by year [3]. Identification of traditional orthopedic images is performed by the orthopedist by manually comparing the image of the mock implant with the X-ray image of the patient before the operation. These methods are the traditional methods of determining the size of a patient's implant. This method used is repeated several times because the manual or observational method takes a long time to determine the dimensions of the patient's implant. Therefore, manual processes should be converted to digital and automatic processes. More specifically, you want to use the software. this system helps surgeons digitally and automatically determine an acceptable implant size for a patient. Several studies have shown that digital techniques can improve placement of total hip arthroplasty. Primary research hip replacement is one of the most common orthopedic surgeries. a total hip replacement removes the damaged hip from arthritis. The ball joint is then replaced with a synthetic implant. The materials used for implants depend on a number of things, including the age of the patient, gender, height, weight, waist and leg length. The importance of hip shape is clearly defined in previous studies 1-5. due to wide variation in the anatomy of the femoral prosthesis, precise seating of the bone implant is difficult to achieve. Asians have a smaller distal femur size than Westerners 6.7. However, the most important artificial femoral prostheses are standardized and manufactured in Europe as well as in the North American region8, and thus the currently available Western orthopedic implants do not match the size of the proximal femur of the Indian population. the use of these oversized and incorrect implants worsens the outcome of surgery, where problems such as stress shielding, tremor and loosening are reported 9-12. This standard hip implant was not useful for the population in the Vidarbha region as it was not supported by anthropometrics for each population13. Dimensional variability may need to be considered when designing an acceptable implant14. To eliminate femur-implant mismatch and achieve proper seating, it was absolutely necessary to modify several standard implants to support the shape and size of the proximal femur of each population. Therefore, the aim of this study was to develop a regular hip implant supporting the anatomical parameters of the

Vidarbha region population. This survey has been conducted since 2014 in the Vidarbha region of central India. 11 anatomical parameters of the femur were identified from the radiographs of 125 patients in the age category of 50-70 years. Of the total number of treated patients, 67 were men and 58 were women. Each radiograph from 125 patients was processed using DICOM Viewer 4.2.1 software. All anatomical parameters of the femur were measured using linear and angular measurement software tools. the exact location of each anatomical parameter whose value was measured is shown in Figure 1. 1 and thus the measured value of each anatomical parameter marked by alphabets in Fig. 1 exhibited in Fig. 2. The following anatomical measurements used for the study: Femoral Head Diameter (FHD): The diameter of the femoral head in the frontal plane Femoral neck diameter (FND): The diameter of the femoral neck in the frontal plane Horizontal Offset (HO): The horizontal distance between the center of the femoral head and also the axis of the shaft in the frontal plane 1437005278130Anatomic parameter on femur Measured value of the anatomical parameter on the femur Vertical offset (VO): The vertical distance between the center of the femoral head and therefore the center of the level of the lesser trochanter Canal width (CW): CW 20 mm above the lesser trochanter on E in the frontal plane of the femur Canal width (CW): CW in the frontal plane, passing through the middle of the lesser trochanter Canal width (CW): CW 20 mm below the lesser trochanter in G in the frontal plane of the femur Channel width (CW): CW 50 mm below the lesser trochanter in H in the frontal plane of the femur Canal width (CW): CW 75 mm below the lesser trochanter in H in the frontal plane of the femur Canal width (CW): CW 100 mm below the lesser trochanter in J in the frontal plane of the femur Neck-Shaft Angle (NSA): The angle between the axis of the shaft and also the axis .

II. LITERATURE SURVEY

This chapter provides a thorough review of the literature on studies reported in the field of total hip arthroplasty. The literature available in the field of hip prostheses can be broadly divided into the following categories: Estimated force on the hip joint.

- Estimation of forces on hip joint.
- Design of hip prosthesis
- Material used for hip prosthesis
- Experimental analysis
- Finite element analysis of hip prosthesis
- Fatigue analysis of hip prosthesis
- Contact and wear analysis of hip prosthesis
- Estimation of forces on hip joint

Artificial hip design and analysis prosthesis It is essential to accurately estimate the force acting. There have been some experimental studies in this direction and some important references are explained in this section. Bergman et al. [1-6] conducted a series of experiments on different types of patients using instrumented hip implants and recorded forces in response to different activities. From their results, they reported that the stress induced in the implant and bone was affected by both the magnitude and direction of the force and the nature of the activity. For some activities, strength has been found to be as high as 870% of body weight (BW). Stansfield and Nicol [8] studied the contact force of two patients with hip prostheses. In their study, the force exerted on the hip joint was calculated using a three-dimensional model of the leg during slow, usually high-speed walking (0.97 to 2.01 m / s). A direct comparison was made between the measured hip contact force and the calculated force. Hurwitz et al. [9] developed an analytical model based on joint kinematics and dynamics to estimate the natural biological fluctuations of muscle strength and its effect on physiologically constrained hip strength.

Artificial hip design

Vora et al. [10] studied the early failure of proximal and non-cemented hip arthroplasty with a follow-up of more than 24 months, and the early failure rates of these prostheses are modern. I found it to be unacceptably high in my hip design. Scifertetal. [11] designed a convexally curved acetabular lip to reduce the tendency for rearrangement. To study the dislocation phenomenon, a 3D nonlinear finite element (FE) model was developed using ABAQUS® software, demonstrating that the new design achieves 28% higher cross-sectional coefficient accumulation during dislocations. Grossetal. performed a finite element analysis of the hollow stem hip prosthesis to reduce femoral stress shielding.

III Material and methods

Patient data

Data of the ten patients (eight male and two female) from the OrthoLoad Hip Joint III database were used in this study. Further information and patient demographics can be found in the following open access publications: Bergmann et al., 2016; Fischer et al., 2018. Osseous landmarks of the femur and pelvis were manually identified by an expert utilizing the software 3D Slicer (https://www.slicer. org) on CT scans of each patient recorded three months after THA. The landmarks are available online at https://orthoload.com/Hip- III-Landmarks.

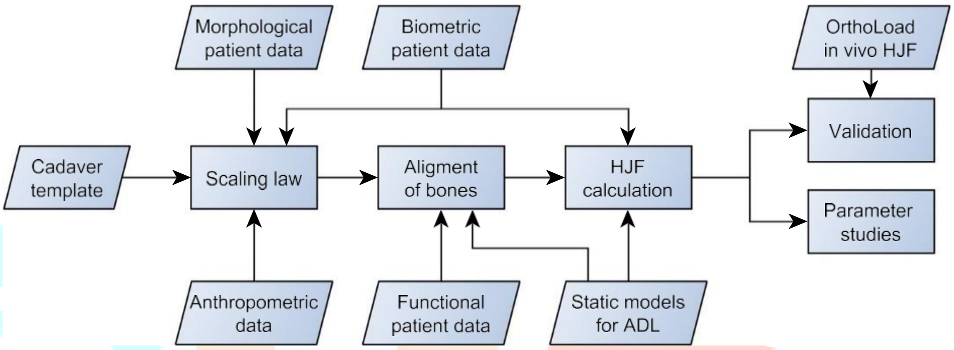
Framework for hip joint force prediction

The framework starts with the selection of a cadaver template (Fig. 1). Bone coordinate systems and joint movements follow the recommendations of the International Society of Biomechanics (Wu et al., 2002). By default, the well-documented and openly accessible TLEM2.0 cadaver template is selected (Carbone et al., 2015). Muscle paths are modeled as straight line, via point and wrapping muscles (Scholz et al., 2016). The cadaver template can be patient-specifically adapted by three different scaling laws. Sub- sequently, the bones of the cadaver template are aligned to a speci- fic body posture based on the static model selected for the corresponding ADL, for instance, the one-leg stance for the normal gait. The static model selected also specifies the method for the HJF calculation and can be directly validated with corresponding in vivo data of the ten Hip Joint III patients. The predicted HJF can be converted from the ISB femoral coordinate system (Wu et al., 2002) to the in vivo HJF presented in the OrthoLoad femoral coordinate system (Bergmann et al., 2016) or vice versa if the land- marks required are available for the cadaver template selected.

Cadaver templates

Three cadaver templates with varying levels of detail were inte- grated into the framework. The Fick1850 cadaver template con- tains 19 hip muscles represented by 44 fascicles modeled as straight lines (Fick, 1850). The Dostal1981 cadaver template con- tains 27 hip muscles represented by 27 fascicles modeled as straight lines (Dostal and Andrews, 1981). The Fick1850 and Dos- tal1981 cadaver templates were considered in this study since they were used by the static models based on Pauwels and Iglic that are described in section 2.5. The TLEM2.0 data set was implemented as the third and most detailed cadaver template. The TLEM2.0 cada- ver template contains 55 hip and leg muscles represented by 166 fascicles (Carbone et al., 2015). Several muscles of the TLEM2.0 cadaver template include via points and wrapping surfaces.

Fig. .1. The framework for hip joint force prediction using static models.



IV MATLAB SIMULATION GUI

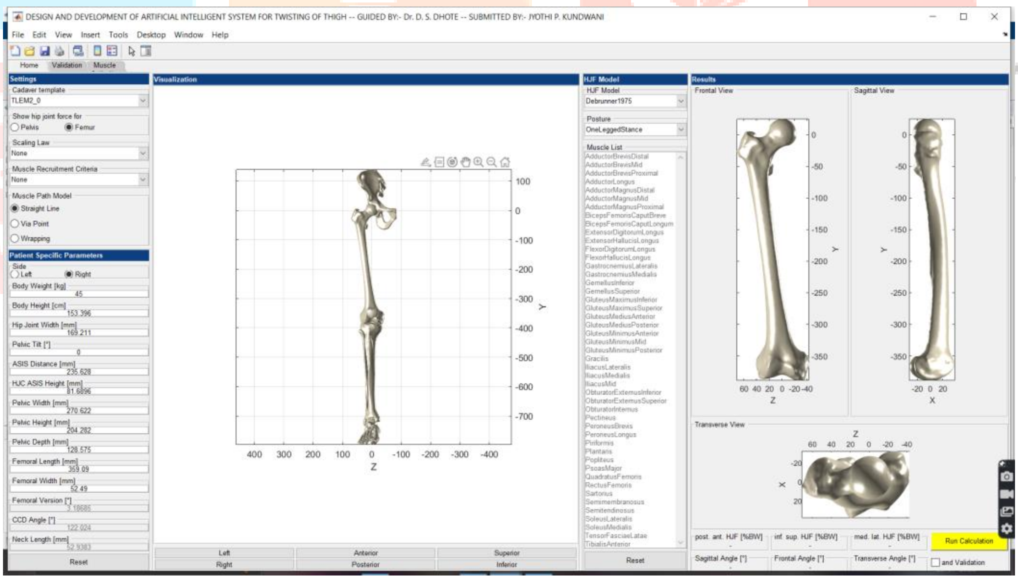


Figure 2 :- GUI of hip implant force

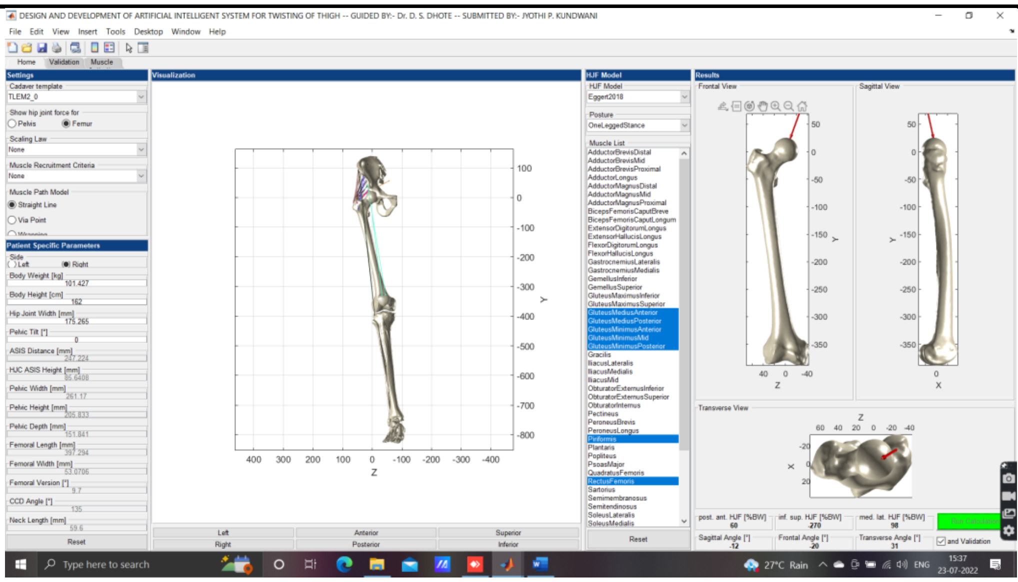


Figure 3:- GUI of hip implant force and muscle selection database

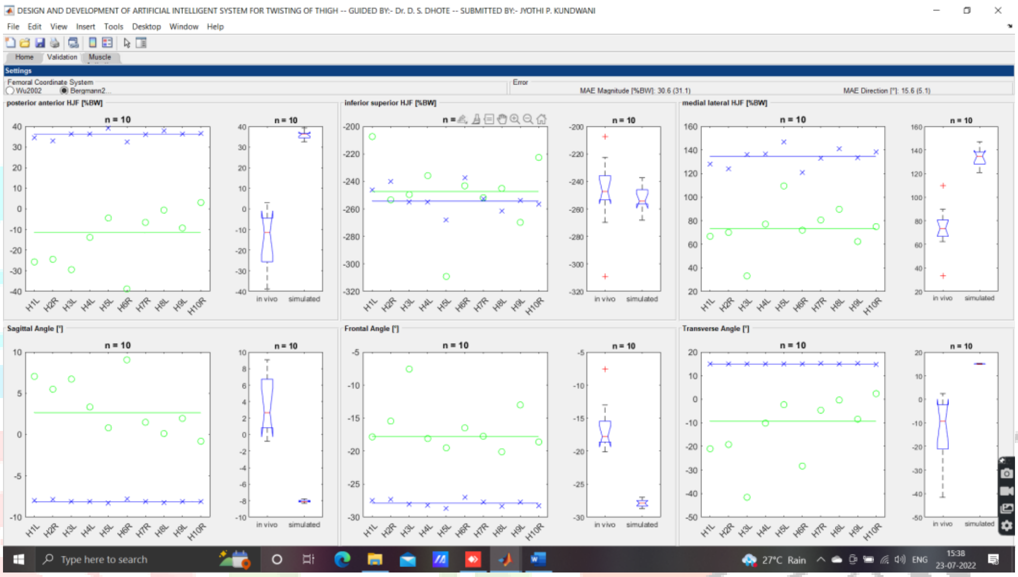


Figure 4:- muscles database validation

V RESULTS OF SIMULATION

Table 1 presents the summary statistics for the comparison of the three unscaled cadaver templates. All results showed a significant error of the predicted HJF in comparison to the in vivo HJF. There was no obvious trend that the error decreases with an increasing level of detail of the cadaver template. Significant differences in magnitude were identified between the cadaver templates for each static model, except for the Iglic model. A combination of the Iglic model and the Fick1850 cadaver was not possible since the muscle grouping of the Iglic model to solve the load sharing problem was not compatible with the Fick1850 cadaver.

Illustrates the effect of the patient-specific scaling on the predicted HJF using the two NUL scaling laws and the Dos- tal1981 cadaver. The LDB scaling law could not be applied to this

Table1:- Median absolute errors (MAE) of the predicted HJF for the comparison of the scaling laws using the Dostal1981 cadaver. The muscles were modeled as straight lines between the origin and insertion.

	Pauwels		Debrunner		Iglic		mediTEC	
Scaling law	MAE Mag.[%BW]	MAE Dir. [°]	MAE Mag. [%BW]	MAE Dir. [°]	MAE Mag. [%BW]	MAE Dir. [°]	MAE Mag. [%BW]	MAE Dir. [°]
One-leg stance								
None	13.2 (21.0)	4.2 (3.1)	24.9 (28.4)	2.8 (3.0)	23.0 (29.4)	3.9 (2.8)	20.1 (24.0)	3.9 (3.5)
NUL$_A$	41.2 (33.6)	3.2 (3.0)	12.6 (20.4)	3.5 (3.6)	16.5 (13.3)	4.4 (5.0)	28.8 (22.9)	4.6 (5.3)
NUL$_B$	12.9 (29.1)	3.8 (2.8)	23.1 (24.2)	3.2 (3.0)	20.1 (20.5)	4.6 (2.4)	18.3 (18.2)	4.7 (2.8)
Level walking								
None	11.9 (27.1)	4.1 (3.3)	22.7 (23.3)	4.3 (2.2)	23.3 (20.9)	4.4 (2.4)	14.6 (5.0)	3.8 (2.2)
NUL$_A$	41.7 (14.5)	2.9 (3.6)	9.1 (19.0)	3.6 (2.1)	11.7 (10.9)	4.1 (2.0)	24.3 (10.8)	4.2 (2.1)
NUL$_B$	17.5 (29.4)	4.0 (3.6)	19.5 (18.8)	4.5 (3.5)	19.9 (19.6)	4.5 (3.8)	12.6 (5.6)	4.1 (2.8)

Table2:- Median absolute errors (MAE) of the predicted HJF for the comparison of the scaling laws using the TLEM2.0 cadaver. The muscles were modeled considering via points and wrapping surfaces

	Pauwels		Debrunner		Iglic		mediTEC	
Scaling law	MAE Mag. [%BW] Dir. [°]	MAE	MAE Mag. [%BW]	MAE Dir. [°]	MAE Mag. [%BW]	MAE Dir. [°]	MAE Mag. [%BW]	MAE Dir. [°]
One-leg stance								
None	34.6 (29.3)	3.8 (2.7)	19.9 (21.5)	3.5 (2.7)	34.8 (28.0)	6.1 (2.5)	111.0 (54.5)	6.8 (2.5)
NUL$_A$	58.8 (27.1)	3.5 (3.0)	18.6 (12.3)	3.5 (3.6)	12.5 (19.8)	7.6 (4.0)	119.5 (25.7)	6.9 (4.8)
NUL$_B$	32.9 (28.3)	3.5 (3.1)	18.4 (20.0)	3.3 (3.2)	27.5 (24.3)	6.7 (2.2)	87.2 (49.9)	6.1 (2.7)
LDB	44.1 (32.4)	3.3 (2.3)	16.4 (19.8)	2.8 (2.3)	15.3 (17.0)	5.6 (5.5)	29.8 (21.1)	4.6 (3.7)
Level walking								
None	40.7 (29.4)	6.5 (3.2)	14.8 (5.0)	6.0 (3.3)	26.6 (28.6)	5.8 (4.9)	118.9 (75.7)	6.7 (3.4)
NUL$_A$	55.7 (12.1) (2.2)	3.8	16.1 (7.1)	3.6 (2.2)	10.7 (8.3)	6.9 (4.7)	130.3 (29.7)	5.5 (4.3)
NUL$_B$	40.3 (29.0)	5.8 (4.6)	13.7 (4.7)	5.3 (4.6)	20.0 (25.4)	6.0 (6.1)	88.7 (39.6)	5.6 (3.7)
LDB	51.5 (13.8) (2.4)	3.7	12.7 (9.8)	3.3 (2.8)	4.4 (21.0)	5.5 (6.5)	38.4 (37.6)	4.9 (5.1)

Table3:- Median absolute errors (MAE) of the predicted HJF for the comparison of different recruitment criteria. The min/max criterion is equivalent to the polynomial criterion of infinite degree (Rasmussen et al., 2001). The muscles were modeled considering via points and wrapping surfaces

		Polynomial 1		Polynomial 2		Polynomial 3		Polynomial 5		Min/Max	
Cadaver	Scaling law	MAE Mag. [% BW]	MAE Dir. [°]	MAE Mag. [% BW]	MAE Dir. [°]	MAE Mag. [% BW]	MAE Dir. [°]	MAE Mag. [% BW]	MAE Dir. [°]	MAE Mag. [% BW]	MAE Dir. [°]
One-leg stance											
TLEM2.0	LDB	14.1 (24.8) (3.7)	4.0	29.8 (21.1)	4.6 (3.7)	50.2 (66.3)	5.1 (3.5)	70.8 (93.7)	5.4 (3.3)	129.2 (115.5) 4.8 (3.7)	
Level walking											
TLEM2.0	LDB	9.7 (13.2) (3.9)	4.0	38.4 (37.6)	4.9 (5.1)	52.7 (40.9)	5.1 (5.6)	86.7 (58.1)	5.3 (6.0)	138.9 (73.6) (6.7)	4.8

cadaver template since the surface data of the bones were missing. Significant errors compared to the in vivo HJF were present for all models and scaling laws. None of the scaling laws improved the predicted HJF significantly in comparison to the unscaled cadaver. Table 3 compares the errors of the predicted HJF using the three different scaling laws and the TLEM2.0 cadaver. Significant errors compared to the in vivo HJF were present for all models and scaling laws. The Iglic model was significantly improved in magnitude by the NUL$_B$ scaling law and the mediTEC model by the NUL$_B$ and LDB scaling law. However, none of the scaling laws improved the predicted HJF significantly in both magnitude and direction.

Table 4 provides an overview of the impact of the muscle recruitment criterion on the predicted HJF using the mediTEC model with the TLEM2.0 cadaver and the LDB scaling law. The error of the predicted HJF in magnitude increases significantly with the increasing degree of the polynomial criterion, with the min/max criterion being equivalent to the polynomial criterion of infinite degree (Rasmussen et al., 2001).

Table 5 presents a more detailed analysis of the prediction error of the three models with a MAE Mag. below 15 %BW and MAE Dir. below 5°. No significant differences between one-leg stance and level walking were found for the AE and error of the three force components.

Fig. 3 exemplarily shows the necessary reduction of the load- based target zone due to the prediction uncertainty of the HJF for the Ortho Load patient H2R. The prediction error for level walking of the Debrunner model, Dostal1981 cadaver and NUL_A scaling law was selected. The results of the Ortho Load patient H2R were almost identical to the median reduction of all ten Ortho Load patients. The latter was 16% (1%) for the MAE, 29% (1%) for the Q3 AE and 53% (2%) for the maximum AE.

Conclusion

Notwithstanding the relatively limited sample size, the study offers valuable insights into the effect of the underlying cadaver template on the prediction of the HJF using static models. The hypothesis that a more detailed patient-specific adaptation of the osseous morphology improves the prediction significantly could only be confirmed partially for one of the cadaver templates. The necessary reduction of the load-based target zone due to the pre- diction uncertainty questions the use within a preoperative planning framework, although the prediction error of the peak HJF of the static models was similar in magnitude and even smaller in direction compared to dynamic models. However, the definition of the edge loading used in this study to calculate the load-based target zone might be too strict and should be further investigated. Future work will include the development of static models for other ADL, such as sit-to-stand or stair climbing, integration of additional cadaver templates and evaluation of more detailed scaling laws also considering the scaling of the PCSA.

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