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Prediction of Kinematic and Kinetic Performance in a Drop Vertical Jump with Individual Anthropometric Factors in Adolescent Female Athletes: Implications for Cadaveric Investigations

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Abstract

Anterior cruciate ligament injuries are common, expensive to repair, and often debilitate athletic careers. Robotic manipulators have evaluated knee ligament biomechanics in cadaveric specimens, but face limitations such as accounting for variation in bony geometry between specimens that may influence dynamic motion pathways. This study examined individual anthropometric measures for significant linear relationships with *in vivo* kinematic and kinetic performance and determined their implications for robotic studies. Anthropometrics and 3D motion during a 31 cm drop vertical jump task were collected in high school female basketball players. Anthropometric measures demonstrated differential statistical significance in linear regression models relative to kinematic variables (P -range < 0.01-0.95). However, none of the anthropometric relationships accounted for clinical variance or provided substantive univariate accuracy needed for clinical prediction algorithms ($r^2 < 0.20$). Mass and BMI demonstrated models that were significant ($P < 0.05$) and predictive ($r^2 > 0.20$) relative to peak flexion moment, peak adduction moment, flexion moment range, abduction moment range, and internal rotation moment range. The current findings indicate that anthropometric measures are less associated with

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CONFLICT OF INTEREST STATEMENT

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kinematics than with kinetics. Relative to the robotic manipulation of cadaveric limbs, the results do not support the need to normalize kinematic rotations relative to specimen dimensions.

Keywords

anterior cruciate ligament; robotic manipulator; knee kinematics; anthropometric variability; cadaveric simulation

INTRODUCTION

In the United States an estimated 250,000 people sustain anterior cruciate ligament (ACL) injuries each year and over 125,000 ACL reconstructions are performed to repair these injuries.¹³ Conservative cost estimates place the corresponding expense to surgically reconstruct and rehabilitate these injuries in excess of \$1 billion annually.¹⁰ Yet despite these costs, the long term outlook of ACL reconstruction is not encouraging. Greater than 50% of patients with ACL reconstructions will experience early onset osteoarthritis within 10 years post-surgery.¹⁶ In studies of both male and female soccer athletes 12-14 years after ACL reconstructions, 75-84% of patients reported deterioration in knee quality of life, which corresponds with the predicted onset of osteoarthritis.^{15,31} Furthermore, patients with ACL reconstructions are far more likely to suffer another ACL injury than healthy controls, as 13% of reconstructed athletes incur a secondary ACL injury, while only 1% of their healthy counterparts incur a primary ACL injury.^{19,26,27} This rate was greater in an athlete-specific population that returned to sport following ACL reconstruction, as epidemiologic data within the first year of return identified secondary injury rate to be as high as 24%.²⁴ One rationale for these bleak post-reconstruction outcomes is that researchers lack detailed knowledge of the functional intra-articular biomechanics of the native ACL and ACL reconstructions during athletic activities. Recently, robotic studies have begun to quantify the biomechanical contributions of both the intact ACL and various ACL reconstructions. These studies utilized anterior tibial translation, internal tibial rotation, and valgus torque at the knee to simulate Lachman's and Pivot-Shift clinical exams. Through these investigations it was shown that though ACL grafts improve joint function compared to the ACL-deficient condition, grafts do not match the mechanical restraint properties demonstrated by the intact ACL.^{8,28} Such disparity in biomechanical function between the reconstructed and intact conditions may be the source of joint degeneration and increased injury rates following ACL reconstruction. Due to these shortcomings in restoring biomechanics, long-term patient satisfaction, and injury rate disparity, ACL reconstructions are currently clinically inferior to intact conditions and encourage research related to the prevention of ligament injuries.

One mechanism that has been used for the evaluation of ACL reconstruction mechanics is robotic-driven *in vitro* investigation.^{3,4,9,12} Such studies allow researchers to garner valuable mechanical data from cadaveric and animal models utilizing invasive methods that would be impossible to execute *in vivo*. While some *in vitro* models use simulated impact³² or the path of least mechanical resistance⁴ to articulate joints, others utilize kinematics recorded from 3D motion systems to define position-controlled joint articulations.^{9,12} In animal models it would be feasible to record *in vivo* kinematics, sacrifice the limb, and then use the

subject-specific kinematics as input to constrain the joint position. However, in human models, this practice is impossible. Therefore, the kinematic input applied to a cadaveric model must be derived from a secondary, living athlete. 3D kinematic reliability has been documented within and between subjects performing the same athletic task.⁷ Between-subject kinematic reliability is lower than within-subject reliability; therefore, the introduction of kinematics recorded from one subject onto a cadaveric limb from a second subject may introduce errors in joint articulation. Due to biologic variability, it is unlikely that a cadaveric specimen and *in vivo* motion subject share identical anatomical geometry. Therefore, it would be useful to understand if differences in kinematic performance could be predicted relative to anthropometric properties such as height and mass. If these associations between basic anthropometric measures and kinematic performance were identified, then *in vivo* kinematics could be scaled relative to the size of each cadaveric specimen prior to their inclusion in simulation models. Any specimen-specific normalization applied to cadaveric simulations is likely to decrease inter-specimen variability and strengthen the power of findings.

The purpose of this study was to examine individual anthropometric measures for significant and clinically predictive linear relationships with kinematic and kinetic performance during a drop vertical jump (DVJ). The hypothesis tested was that anthropometric measures would not impact the magnitude of kinematic joint rotations observed between subjects, but would impact kinetics.

METHODS

Participants in the current study consisted of a cohort of 239 middle and high school female basketball athletes (mass = 55.4 ± 13.2 kg, height = 1.60 ± 0.09 m, tibia length = 0.31 ± 0.03 m, BMI = 21.3 ± 3.9 , age = 13.6 ± 1.6 years) from a prospective, longitudinal study. Female athletes were selected as the study population because they experience ACL injuries at 4 to 6 times the rate of their male counterparts.¹¹ Testing procedures were approved by the institutional review board and informed, written consent was obtained from the parent or legal guardian of each subject. Each subject also provided consent prior to participation.

Participants were evaluated for anthropometric measures prior to motion testing. A stadiometer was used to measure height with subjects standing barefoot. A calibrated physician's scale was used to measure body mass again with subjects standing barefoot. Participants were also measured for shoe size as footwear for motion testing was provided to them.

Subjects were instrumented with 43 retro-reflective markers for 3D biomechanical analysis. Markers were arranged in a modified Helen Hayes format that has been previously described.² Motion data was collected and sampled at 240 Hz with a 10 camera motion analysis system (Eagle cameras, Motion Analysis Corporation, Santa Rosa, CA). Ground reaction forces (GRF) were collected by dual, in-ground, multi-axis force platforms (BP600900, AMTI, Watertown, MA) and sampled at 1200 Hz. Prior to dynamic motion testing a static standing trial was collected for each subject to define body segments,

dimensions, and neutral alignment. All joint angles were reported in reference to this neutral alignment.

Each participant performed three DVJ trials starting from a 31 cm box.^{2,7} Motion was recorded for each trial and all successful trials from a subject were averaged into an individual mean. A trial was deemed successful if the subject left the initial box simultaneously with both feet and landed on the force platforms simultaneously with each foot entirely contained within separate plates. Trials that did not match this criteria were excluded and the subject average was taken from the remaining trials. Contact phase motion data were processed in Visual3D (version 4.0, C-Motion, Inc., Germantown, MD) with custom MATLAB code (version 2012a, The Mathworks, Inc, Natick, MA). Visual3D used the relative positions of retroreflective markers to define each body segment as a rigid body with length and volume, while an internal biomechanical model assigned segment mass based on a percentage of the subject's bodyweight. Data processing methods for segment definition as well as joint angle calculation followed previously specified conventions.⁷ Only the first contact phase of the DVJ was analyzed. Contact phase was identified with GRF data and defined as the point of initial contact (IC) through toe-off. IC was defined as the first point where the ground reaction force exceeded 10 N, whereas toe-off was defined as the first point after IC where the ground reaction force was below 10 N. Marker trajectories and GRF data were processed through a fourth-order, low-pass, digital filter with a cutoff frequency of 12 Hz for kinetic and kinematic calculations. All moments were reported as external joint moments derived from the ground reaction forces created during contact with the force platforms.

From the data collection process, four anthropometric (height, weight, body mass index (BMI), and tibia length) measures were selected as independent variables. These variables were selected for analysis as they represent easily identifiable anthropometrics that have previously been incorporated in multi-factorial ACL injury risk assessments.^{21,22} Variables were also selected relative to the anthropometric data that is typically included with cadaveric specimen procurement. Twelve kinematic (knee flexion, abduction, and internal angle at IC; maximum abduction, adduction, internal, external, flexion, and minimum flexion angle; and flexion, abduction, and internal rotation ranges of motion (ROM)) and 12 kinetic (knee flexion, abduction, and internal moment at IC; maximum abduction, adduction, internal, external, flexion, and minimum flexion moment; and flexion, abduction, and internal moment ROM) knee measures were then selected as dependent variables. Peak values and ROMs were selected as dependent variables for each rotational degree of freedom as metrics of this nature are associated with and used to predict ACL injury risk; and therefore, should also be relevant to robotic simulations that wish to model ACL loading.^{11,21} Each of the four independent variables was individually correlated with each of the 24 dependent variables, through single-factor linear regression models. An alpha level of 0.05 was used to judge statistical significance in all models. A 0.2 linear regression r^2 cutoff criterion was then used to determine if a significant model was correlated enough to be considered predictive. This cutoff criterion was previously established as a requirement for variable inclusion in multivariate linear models used to predict ACL injury risk from kinematic and kinetic variable data.¹¹ A backward elimination technique found that variables with $r^2 < 0.2$ did not have a significant impact on the predictive model and should

be excluded. The joint and motion task examined in the present study were the same as previous literature; therefore, the predictive threshold was maintained. Following univariate linear regression, these same criteria were used to evaluate multivariate linear regression models that correlate all four independent variables collectively against each of the 24 dependent variables. All statistics were performed in MATLAB with the Statistical Toolbox and custom code and verified through SPSS.

RESULTS

Kinematics

Linear model relationships between individual anthropometric independent variables and kinematic dependent variables were significant ($P < 0.05$) in 27 out of the 48 variable combinations (56%). The height, BMI, and tibia length variables produced the greatest number of significant models relative to kinematic variables at seven each. Of the 26 significant linear models, no model reached the 0.2 linear regression r^2 cutoff criterion to be considered a predictive model (Table 1). The largest r^2 for kinematic models was mass correlated with minimum flexion angle ($r^2 = 0.17$; Figure 1). All but three of the significant linear models, which were all relative to maximum flexion angle, displayed an $r^2 < 0.10$. Linear relationships were significant in all multivariate models apart from flexion angle at IC and flexion angle ROM. For all 24 dependent kinematic variables, the multivariate models universally exhibited larger r^2 values than the corresponding univariate models. However, the only model that reached the predictive cutoff criterion was for minimum flexion angle ($r^2 = 0.20$).

Kinetics

Linear model relationships between individual anthropometric independent variables and kinetic dependent variables were significant ($P < 0.05$) in 39 out of 48 cases (81%). Height, mass, and BMI produced the greatest number of significant models relative to kinetic variables with 10 each. Of the 39 significant models, 16 exceeded the 0.2 linear regression r^2 cutoff criterion to be considered a clinically predictive model (Table 2). The predictive linear models were height, mass, BMI, and tibia length relative to maximum flexion moment; mass and BMI relative to maximum extension moment; height, mass, BMI, and tibia length relative to the range of flexion moment values; height, mass, and BMI relative to the range of abduction moment values; as well as height, mass, and BMI relative to the range of internal moment values. The most predictive linear models were mass correlated with the knee flexion moment range and maximum flexion moment values ($r^2 = 0.63$ and 0.68 , respectively; Figure 2). BMI correlated with knee flexion moment range and maximum flexion moment had $r^2 > 0.44$, whereas the remaining predictive models demonstrated r^2 values between 0.22 and 0.42. The rates of change for all between independent and dependent variables for all predictive models are documented in Table 3. As with the kinematic dependent variables, multivariate regression models exhibited greater r^2 values than univariate models for all 24 kinetic variables. However, in all cases these increases were only marginally larger than the univariate model based on mass (r^2 difference = 0.04). The multivariate models for maximum knee flexion moment, maximum knee extension

moment, flexion moment range, abduction moment range, and internal moment range all exceeded the predictive r^2 threshold.

DISCUSSION

The objective of the current study was to examine anthropometric measures for significant and clinically-predictive linear relationships with kinematic and kinetic performance during a DVJ. It was believed that the identification of these linear correlations would provide a baseline by which to scale *in vivo* recorded kinematics relative to the geometry of each cadaveric specimen. Such specimen-specific considerations would optimize robotic simulation models through a likely reduction in variability.

Though 65 of the individual linear models examined were found to represent significant relationships between their respective independent and dependent variables, poor r^2 values indicated that most of these associations did not account for enough of the variance to serve as clinically predictive models. Our lab has previously documented that in order for a variable to be considered predictive of kinematic and kinetic outcomes the coefficient of determination must account for greater than 20% of the error ($r^2 > 0.2$).¹¹ Congruent with this range of significance, Fleming et al⁵ concluded that his isometer model could not accurately predict ACL reconstruction tension from an r^2 value of 0.15. Of the 65 significant linear models found in the present study, 49 fell below the 0.2 threshold and therefore were not considered clinically predictive. No anthropometric variables were found to be clinically predictive of any kinematic factors. As anthropometric measures were not able to accurately predict changes in kinematics, it may be inappropriate to adjust rotational kinematic input relative to specimen size in cadaveric studies.

Though 16 linear models exceeded the necessary threshold to be considered predictive, the presently reported r^2 values were below what is typically observed in the literature. Many documented clinical predictors of various knee behaviors account for greater than 70% of the variance in the model. Prior reports predicted ACL injury with $r^2 = 0.88$,¹¹ while knee joint forces after total knee replacement have been predicted with r^2 between 0.77-0.85,¹⁷ and lower leg motion in response to electrical stimulation has been predicted with r^2 between 0.71-0.94.²⁵ As none of the predictive linear models identified in this study exceeded $r^2 = 0.70$, they should be considered moderate to weak predictors. The presented models' inability to account for variance in prediction may be a result of their single independent variable format. Previously developed ACL injury risk prediction nomograms rely upon five^{21,22} and 15³⁰ variables to derive their calculations with high sensitivity and specificity. Therefore, the weak r^2 values in the current study may indicate that models based on single independent variables may be too oversimplified to accurately predict kinematic and kinetic values. Similarly, the presented anthropometric multivariate regression models also failed to be significantly predictive of kinematic performance. In multivariate models used to predict ACL injury risk for adolescent athletes, additional intrinsic variables, such as hamstrings:quadriceps activation ratio, not based on anthropometric measures are incorporated to increase predictive accuracy.²¹ Though hamstrings:quadriceps strength ratio might contribute significantly to kinematic prediction,

as it does to ACL injury risk prediction, that data cannot be obtained from a cadaveric specimen and would be useless to the optimization of robotically-driven joint simulations.

Moving forward, this research indicated that kinematic rotational inputs for 6-degree-of-freedom robotic simulations of cadaveric knee joint articulation should not be scaled relative to specimen-size. This point will serve as an important justification relative to the design of kinematic inputs during the development of reliable and reproducible robotic simulations with the capacity to recreate *in vivo* athletic tasks on cadaveric knees. One such simulation model that will be utilized to examine intra-articular knee biomechanics relative to ACL injury risk during drop vertical jump and sidestep cutting tasks is currently under development by the authors.

Unfortunately, the linear models that demonstrated both significant linear relationships and moderate predictive abilities failed to correspond with the kinematic and kinetic knee variables associated with ACL injury risk. Knee abduction angle and moments^{6,11,22} and internal tibial rotation^{20,29} have been associated with or used to predict likelihood of ACL injury. The moderately predictive models in this study were only relative to peak and minimum flexion moment, range of flexion moment values, range of abduction moment values, and range of internal moment values. Sagittal plane moments were the most predictive models in the present study, but previous investigations have identified that sagittal plane moments alone do not threaten ACL integrity during the performance of athletic tasks.¹⁸ As clinical investigations have not identified such variables as key contributors to ACL injury, the moderately predictive linear models identified in the current study do not offer much benefit for the optimization of robotically-driven cadaveric knee models relative to ACL biomechanics.

Independent anthropometric variables were found to have more significant associations with kinetics than kinematics. The relative significance of these variables to kinetic prediction adheres to mechanical principles as moments are the product of force and distance. An extended tibia length should rapidly increase knee moments as it would move the 2.0-2.5 times bodyweight GRFs incurred on a single leg when landing from a DVJ further away from the knee.² Similarly, increased subject mass will also increase the raw values of landing GRFs as they are relative to bodyweight. This finding echoes the weighted structure of the previously reported ACL injury risk prediction nomogram.^{21,22} In that nomogram, tibial length has the highest weight and can account for up to 100 points of the total injury prediction score for each subject. A high tibial length score alone can place an athlete with a 0.75 probability of high knee moment.

Pennation angles and lines of action for the major muscle groups surrounding the knee primarily correspond to the generation of flexion/extension torque and anterior/posterior loading at the joint.^{14,33} As such, during neuromuscular-controlled *in vivo* tasks, the sagittal plane has been shown to experience a greater range of articulation and torque as it stabilizes against a greater proportion of ground reaction force than any other plane of motion at the knee.¹ Therefore, it is unsurprising that linear regression models developed from anthropometric measures, especially mass, correlated best with peak sagittal plane torques. Larger mass should increase the mechanical demand at the knee. Unless there is a loss of

neuromuscular control, load increases should mostly be countered through sagittal plane loading which leads to prevalent flexion/extension correlations. Correspondingly, the frontal and transverse planes, which support a lower proportion of knee loading,¹ were less correlated with changes in anthropometric variables.

One limitation of the current study is the age disparity between *in vivo* subjects and cadaveric specimens. It is known that the mechanical properties of tissue decline with age.²³ However, to obtain cadaver specimens with an average age of 13.6 that matches the *in vivo* population would be unreasonably difficult. The decline in mechanical properties with older cadaveric specimens relative to our *in vivo* population likely would affect absolute peak loading magnitudes, but still exhibit the same trends of change. Therefore, the conclusions drawn in this project should still be applicable to cadaveric populations.

In conclusion, both univariate and multivariate linear models derived from simple anthropometric variables may not be sufficiently robust to be instituted as clinical predictors of kinematic performance during a DVJ. More complex multivariate nomograms, that incorporate non-anthropometrically based variables, are necessary to accurately predict kinematic behaviors. Relative to cadaveric-based orthopaedic research, the current findings indicate that kinematics are not correlated with stature, and therefore provide no current evidence that input kinematic rotations should be normalized to specimen anthropometrics during simulations.

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Figure 1.

Data plot of subject mass versus minimum knee flexion angle. This plot exemplifies data where the factors exhibited a significant linear relationship, but a poor r^2 correlation.

Figure 2.

Data plot of subject mass versus knee flexion moment range. This plot exemplifies data where the factors exhibited a significant linear relationship and a moderately strong r^2 correlation.

Table 1

Depicts the r^2 values for linear relationships between the four independent variables (columns) and 12 kinematic dependent variables (rows).

Dependent Variable (°)	Height	Mass	BMI	Tibia Length
Flexion Angle @ IC	0.00	0.03*	0.03*	0.00
Abduction Angle @ IC	0.06*	0.00	0.01	0.05*
Internal Angle @ IC	0.04*	0.04*	0.02*	0.05*
Max Flexion Angle	0.06*	0.08*	0.06*	0.02*
Min Flexion Angle	0.12*	0.17*	0.12*	0.02*
Max Abduction Angle	0.09*	0.00	0.01	0.08*
Max Adduction Angle	0.06*	0.00	0.03*	0.05*
Max Internal Angle	0.00	0.04*	0.05*	0.01
Max External Angle	0.02	0.04*	0.03*	0.01
Flexion Angle ROM	0.00	0.00	0.00	0.02
Abduction Angle ROM	0.06*	0.01	0.00	0.06*
Internal Angle ROM	0.02	0.00	0.01	0.00

* Indicates that the linear regression model between the independent and dependent variable was significant ($P < 0.05$)

Table 2

Depicts the r^2 values for linear relationships between the four independent variables (columns) and 12 kinetic dependent variables (rows).

Dependent Variable (N*m)	Height	Mass	BMI	Tibia Length
Flexion moment @ IC	0.02 *	0.07 *	0.06 *	0.01
Abduction moment @ IC	0.00	0.00	0.00	0.00
Internal moment @IC	0.00	0.00	0.00	0.00
Max Flexion Moment	0.42 *	0.63 *	0.44 *	0.23 *
Max Extension Moment	0.12 *	0.32 *	0.27 *	0.04 *
Max Abduction Moment	0.09 *	0.07 *	0.03 *	0.06 *
Max Adduction Moment	0.07 *	0.17 *	0.15 *	0.05 *
Max Internal Moment	0.10 *	0.17 *	0.12 *	0.05 *
Max External Moment	0.08 *	0.05 *	0.02 *	0.05 *
Flexion Moment Range	0.42 *	0.68 *	0.48 *	0.22 *
Abduction Moment Range	0.27 *	0.38 *	0.26 *	0.19 *
Internal Moment Range	0.25 *	0.27 *	0.15 *	0.14 *

* Indicates that the linear regression model between the independent and dependent variable was significant ($P < 0.05$)

Table 3

Depicts the slope of the best fit line (rate of change) between independent and dependent variables that were found to have both a significant linear relationship and a $r^2 > 0.20$.

Dependent Variable	Independent Variable	Rate of Change
Max Flexion Moment	Height	$-2.234 \text{ N m cm}^{-1}$
Max Flexion Moment	Mass	$-1.837 \text{ Nm kg}^{-1}$
Max Flexion Moment	BMI	$-5.106 \text{ N m BMI}^{-1}$
Max Flexion Moment	Tibia Length	$-0.5515 \text{ N m cm}^{-1}$
Max Extension Moment	Mass	$0.253 \text{ N m kg}^{-1}$
Max Extension Moment	BMI	$0.776 \text{ N m BMI}^{-1}$
Flexion Moment Range	Height	$2.463 \text{ N m cm}^{-1}$
Flexion Moment Range	Mass	$2.090 \text{ N m kg}^{-1}$
Flexion Moment Range	BMI	$5.882 \text{ N m BMI}^{-1}$
Flexion Moment Range	Tibia Length	$0.596 \text{ N m cm}^{-1}$
Abduction Moment Range	Height	$0.627 \text{ N m cm}^{-1}$
Abduction Moment Range	Mass	$0.499 \text{ N m kg}^{-1}$
Abduction Moment Range	BMI	$1.370 \text{ N m BMI}^{-1}$
Internal Moment Range	Height	$0.208 \text{ N m cm}^{-1}$
Internal Moment Range	Mass	$0.144 \text{ N m kg}^{-1}$