Sports Eng
DOI 10.1007/s12283-008-0005-0

ORIGINAL ARTICLE

Segmental sequencing of kinetic energy in a computer-simulated golf swing

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Abstract The concept of the transfer of kinetic energy (KE) sequentially through the human body from proximal to distal segments is an influential concept in biomechanics literature. The present study develops this area of research through investigation of segmental sequencing of the transfer of KE by means of computer simulation. Using a musculoskeletal computer model previously developed by the authors, driven using three-dimensional kinematic data from a single elite male golfer, combined inverse and forward dynamics analyses enabled derivation of KE. Rigid body segments of torso, hips, arms and clubhead were examined in line with previous literature. Using this method a driver swing was compared to a 7 iron swing. Findings showed a high level of correlation between driver and iron peak KE and timing of peak KE relative to impact. This seems to indicate equivalent trunk and arms linear velocity, thus force applied, for an iron shot and a driver shot. There were highly significant differences between KE output for body segments for both clubs. In addition, peak KE magnitudes increased sequentially from proximal to distal segments during swing simulations for both the driver and 7 iron. This supports the principle of the summation of speed. However, timing of peak KE was not sequential from proximal to distal segments, nor did segments peak simultaneously. Rather, arms peaked first, followed by hips, torso and club. This seems to indicate a subjective optimal coordination of sequencing.

Keywords Computer simulation · Golf · Kinetic energy · Segmental sequencing

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1 Introduction

Sports activities including overarm throwing, shot putt, discus and the golf swing involve reliable sequencing of body segment movements to produce optimal motion patterns and to achieve said activity's goal. Proximal to distal segmental sequencing is one principle that has been widely used in the literature (including [18]) to describe the kinematics of such activities. This principle suggests that optimum distal segment velocity is achieved via sequential peak velocity from proximal to distal body segments.

Proximal to distal sequencing, or the summation of speed principle as it is often termed, for the golf swing, begins with the onset of the downswing whereby the hips lead the torso, which in turn leads the rotation of the arms followed by the club being swung. A reversed 'C' position is achieved at the end of the follow-through from a combination of torso, hip and arm rotations resulting in weight shift patterns which transfer the majority of the golfer's weight onto the back foot during the backswing and the front foot as the downswing progresses.

Work has been presented, though, in opposition of the summation of speed principle, suggesting an optimal coordination of segments to produce greatest peak clubhead velocity for the golf swing, which may not necessarily produce proximal to distal peak velocity sequencing [17, 18]. Instead, angular velocities of the links between segments may peak simultaneously [22], or in an order specific to an individual. The application of the principle of summation of speed and optimum coordination of kinetic energy (KE) transfer to the analysis of the golf swing is not new, having been experimentally and mathematically modelled prevously [2].

This paper presents similar findings to that of Anderson et al. [2] but using different methods. In that study, three-dimensional motion analysis of a drive swing was performed on 45 male scratch golfers and kinematic data fitted to a four-segment mathematical model. They found that peak magnitudes of total KE increased sequentially from proximal to distal segments, while timing of the peaks did not follow a sequential pattern. For the current study, a 19 segment full-body musculoskeletal human model comprising 42 degrees-of-freedom and 111 muscles was generated [12, 13]. The model, validated using novel methods (see Sect. 2.2), accurately represented the drive swing kinematics for a single elite male subject (+1 handicap) and was used to predict the golf swing due to changes in golf club parameters. Few researchers have developed a full-body model of the golf swing, despite recent calls by biomechanists for the development of human equipment modelling [10] and none to date have applied such modelling techniques to investigate the influential concept in the literature of segmental sequencing characterisation.

As purported by Anderson et al. [2] the purpose of this study was to investigate the transfer of speed through the golfer. This is achieved using model data predicted from forward dynamics simulations where KE of body segments were the outcome measure.

2 Method

The current study used one category 1 golfer to represent the golf swing of an elite golfer (single-subject analysis, [21]. Indeed, Hatze [11] stressed the importance and need for subject-specific models to be developed, examining specific segmental inertial properties and parameters rather than scaled models based on population averages. The subject (25 years, 1.80 m, 91.3 kg, +1 handicap) signed an informed consent approved by the University of Ulster and completed an activity and medical history questionnaire.

2.1 Experimental procedures

Full-body motion data during the golf swing were captured at 240 Hz, using a 5-camera MAC^{TM} Falcon Analogue motion analysis system. Image verification was carried out as described by the MAC^{TM} Falcon instruction manual, and the camera system indicated a maximum residual error of 2 mm for each camera. The calibrated volume (3 m × 3 m × 2.5 m high) was greater than that exhibited by the motion of the markers during the swing. An adaptation of Mitchell et al.'s 26 marker setup was used to describe the motion of the golf swing. The present study located 4 wand markers at the arm mid segments, rather than using Mitchell et al.'s head, waist and thoracic spine markers. Reflective passive surface markers were used, eight of which described arm motion and were 1/2 in. (0.0127 m) in diameter, and the remaining 18 markers were 3/4 in. (0.01905 m) in diameter. Figure 1 shows a not-to-scale diagram of the positioning of the 26 surface markers. Segments were described using wand markers (tibial, femoral, humeral and radial wands). Four-inch (0.1016 m) wands were used for the lower extremity whilst 2.5-in. (0.0635 m) wands were used on the upper extremity to: (1) reduce marker vibration and, thus, noise during the high velocity movement of the arms during the swing, and (2) to ensure a comfortable swing for the golfer. In addition to the 26 body markers and 1 shaft marker the MAC^{TM} computer stick model and environment included a further 16 markers to aid model validation (Table 1).

After performing his usual pre-game warm-up the subject hit eight shots with his own driver (200 g 350 cc clubhead, 'stiff' 45 in. shaft) then his own 7 iron. For each shot the MAC^{TM} system tracked the complete swing, and an investigator recorded any anecdotal information relating to the quality of the shot offered by the subject. The subject was instructed to aim along a target line into netting

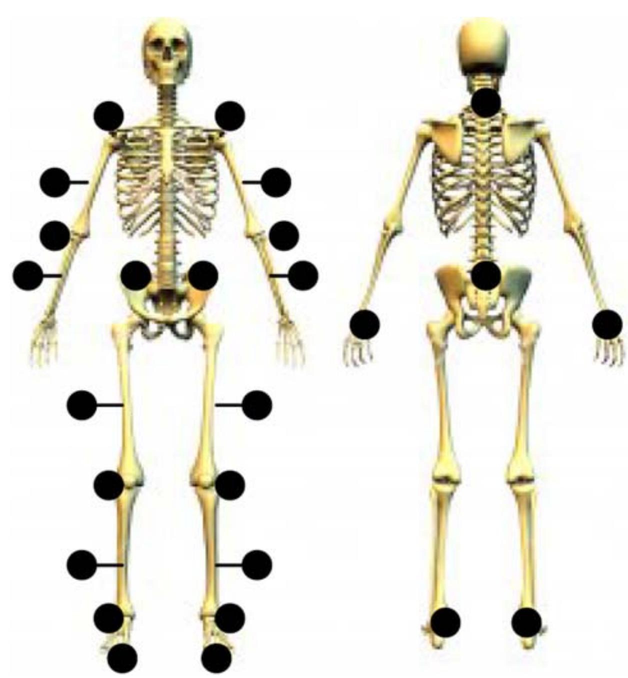


Fig. 1 Diagram showing positioning of the 26 main passive surface markers

Table 1 Additional clone and actual markers used to validate the model

Left greater trochanter	Right acromion clone
Right greater trochanter	Left acromion clone
Left thigh (posterior)	L5 clone
Right thigh (posterior)	Left knee clone
Left inferior patella	Right knee clone
Right inferior patella	Navel
Left medial malleolus	Left ankle_jc
Right medial malleolus	Right ankle_jc

hanging 4.5 m away. Premium golf balls were used for the test.

Reconstructed co-ordinates of the markers to infer joint centre location and segment centre of mass (C.O.M.) were transferred from the capture software MAC EvaRT™ to KinTrak. Data were smoothed using a 12 Hz low-pass second-order Butterworth filter [15]. Data were analysed using SPSS with variables (body segment angular velocity, angles) tested for variance, when different clubs were used, by means of a one-way ANOVA. Variance testing allowed for a trial representative of the series to be selected to drive the computer simulation. Pearson's test for correlation was applied during the validation procedure [13]. A significance level of $P < 0.05$ was set.

2.2 Modelling techniques

The base segment set comprised 19 segments: head, neck, upper torso, central torso, lower torso, right/left scapula, right/left upper arm, right/left lower arm, right/left hand, right/left upper leg, right/left lower leg and right/left foot. The model was initially scaled for the subject's height and body mass (1.80 m, 91.3 kg), with an additional 54 subject-specific measures input detailing such information as segment length, breadth and circumference.

The model in the present study was constructed with a total of 42 degrees-of-freedom. MAC™ kinematic data were used to drive the model inverse dynamics simulations and forward dynamics simulations, which replicated the swing for both driver and iron models. Trainable passive joints for inverse dynamics analysis were created. These joints consisted of a torsional spring force with user-specified stiffness, damping angular limits and limit stiffness values. A full-body set of 111 muscles was generated. Muscles as well as 'soft-tissue' ligament elements transmitted tension forces only. Ligaments were passive spring/dampers and muscle-tendon forces consisted of 'training' elements for inverse dynamics simulations and active (contractile) elements for forward dynamics simulations. The muscle actuators were bound not to exceed the physiological capability of the individual muscle that is constrained by a simple cut off min./max. algorithm. Physiological properties for each muscle included: physiological cross sectional area (pCSA), maximum tissue stress (M_{stress}) which at no point was reached during simulations, and resting load ($F_{resting}$). Contact forces between the body segments and the environment or objects (ground and club grip) were created using ellipsoid-plate contact elements. Modelled driver and iron accurately represented physical properties of the subject's own clubs including material, mass, volume and loft (Table 2).

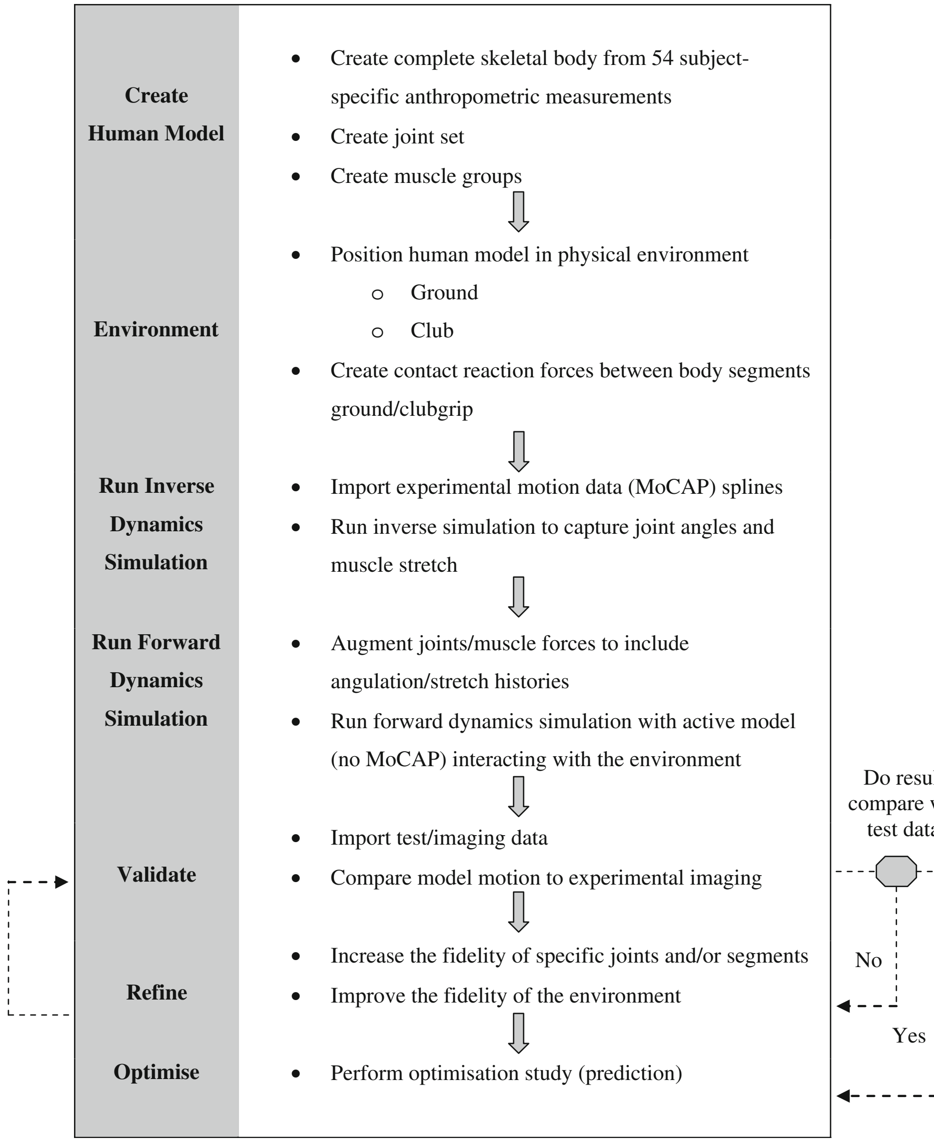
Inverse dynamics simulations were performed on golfer/driver and golfer/iron models driven using respective experimental imaging motion capture data (MoCAP). During these simulations trainable passive joints recorded actual joint angulations. Similarly, the muscles learned shortening/lengthening patterns during the inverse dynamics simulations. The models were then forward simulated, with MoCAP removed, respective golfer/driver and golfer/iron models driven with 'trained driver' joints which held the previous inverse dynamics angulation information and trained muscles with learned activation patterns. The forward dynamics applied calculated joint torque from the recorded joint angulation history, to simulate movement via kinetics. Figure 2 describes the process undertaken to construct the model, apply the kinematic data and process the inverse and forward dynamics simulations.

Table 2 Actual and modelled clubs physical characteristics

Characteristics	Driver		7 iron	
	Actual	Model	Actual	Model
Club length (in. m^{-1})	45/1.14	45/1.14	37/0.94	37/0.94
Shaft material	Carbon composite	Carbon composite[a]	Steel	Steel[a]
Head material	Titanium	Titanium[a]	Steel	Steel[a]
Shaft mass (g)	63.0	63.0	72.0	72.0
Grip mass (g)	51.0	51.0	51.0	51.0
Head mass (g)	200.9	200.9	199.8	199.8
Head volume (cc)	350	350	NA	NA
Loft (°)	9.0	9.0	33.0	33.0

[a] Defined using (a) typical density, (b)Young's modulus and (c) Poisson's ratio

Fig. 2 Flow diagram demonstrating model creation and data processing



2.3 Data analysis

Model validation is normally achieved by comparison of model predicted results with those obtained experimentally for the same condition. For the present study model validation was carried out for velocity, kinematics and kinetics experimental imaging data compared to model data using correlation and root mean square (RMS) statistics.

(a) *Velocity* Experimentally determined peak clubhead C.O.M. velocity was compared to the same measure predicted by the model.
(b) *Kinematics* Additional markers (actual and clone, Table 1) were tracked during experimentation. Such 'validation' markers were not used to drive the model. These markers were replicated virtually on the model and their trajectories recorded during forward dynamics simulations.
(c) *Kinetics* Determination of correct muscle force simulation was achieved by comparison of grip force by the model to grip force reported experimentally in previous research. Grip force was deemed a valid measure of the predicted force produced during the golf swing. Ensuring that the force exerted by the arm muscles and applied to the club compared favourably with previously reported experimental force transducer research meant that reliable simulations had been performed [19].

As in the paper by Anderson et al. [2] the present study examined the KE of four segments of the authors' model, constructed as follows:

1. Club: club head
2. Hips: left foot, right foot, left lower leg, left upper leg, right upper leg, right lower leg.

Fig. 3 Base 19-segment stick model with tri-axis joints

3. Arms: left scapula, right scapula, left upper arm, right upper arm, left lower arm, right lower arm, left hand, right hand.
4. Torso: neck, head, upper torso, central torso, lower torso.

Segments were considered rigid and values for KE[1] were derived from the point of the C.O.M of each combined segment. Figure 3 illustrates the 19 aforementioned basic segments. Data analysed pertained to timing of the swing for both the driver and 7 iron simulation, peak KE of each body segment and the time at which peak KE was produced.

3 Results

3.1 Model validation

Descriptive statistics for model and experimental peak clubhead velocity for each drive length are shown in Table 3. Figure 4 additionally shows the degree of fit between clubhead velocity for an experimental driver trial and a simulated trial. Correlation between experimental and model data was 0.999 with RMS of 1.93 m s^{-1}.

Table 3 Mean (±SD) model and experimental peak clubhead velocity

Club	Model clubhead velocity (m s^{-1})*	Experimental clubhead velocity (m s^{-1})*
Driver	48.302 ± 0.005	50.292 ± 0.632
Iron	33.800 ± 0.006	35.193 ± 0.333

* Pearson's correlation $r = 0.999$, $P < 0.05$; RMS 1.93 m s^{-1}

Table 4 details correlation scores and RMS difference for analysis between a representative selection of experimental validation markers three-dimensional trajectory and its equivalent model predicted values. Correlation was statistically significant ($P < 0.001$) and was strong for each of the 12 markers, averaging 0.983. Root mean square difference averaged 0.05°. Thus, model kinematics can be said to very closely match actual swing kinematics.

The final method of validation concerned grip force, as a measure of the ability of the model to correctly predict muscle force output during the swing. Table 5 compares model data against two sources and shows that model data lie within the range reported by both. It should be noted that, naturally, grip force will vary between golfers, depending on personal grip preference and swing speed. However, it was deemed important to ensure that muscle force predicted by the model in the present study was representative of actual force determined experimentally.

3.2 Kinetic energy

The sequence of peak KE for the single elite subject using a driver during one trial is shown in Fig. 5. Table 6 shows the values of peak KE and its relative timing to impact for the driver trial. For this trial, the magnitude of the KE peak increased starting proximally at the hips and moving distally to the club. As Fig. 6 and Table 7 also show for the 7 iron, the arms peaked first followed in a sequential pattern of the hips, torso then clubhead, all prior to impact.

The sequence of peak KE for the single elite subject using a 7 iron during one trial is shown in Fig. 6. Table 7 shows the values of peak KE and its relative timing to impact for the 7 iron trial.

Peak KE for the clubhead was found to be just 79% that predicted for the same segment during driver simulations. Again, the magnitude of the KE peak increased starting proximally at the hips and moving distally to the club. Interestingly though, the timing sequence remained the same as the driver trial, and with nearly identical relative timing of peak KE to impact by all segments except the arms.

[1] KE = ½I $\dot{\theta}^2$ where $\dot{\theta}$ is the tangential linear velocity of the C.O.M. of the respective segment.

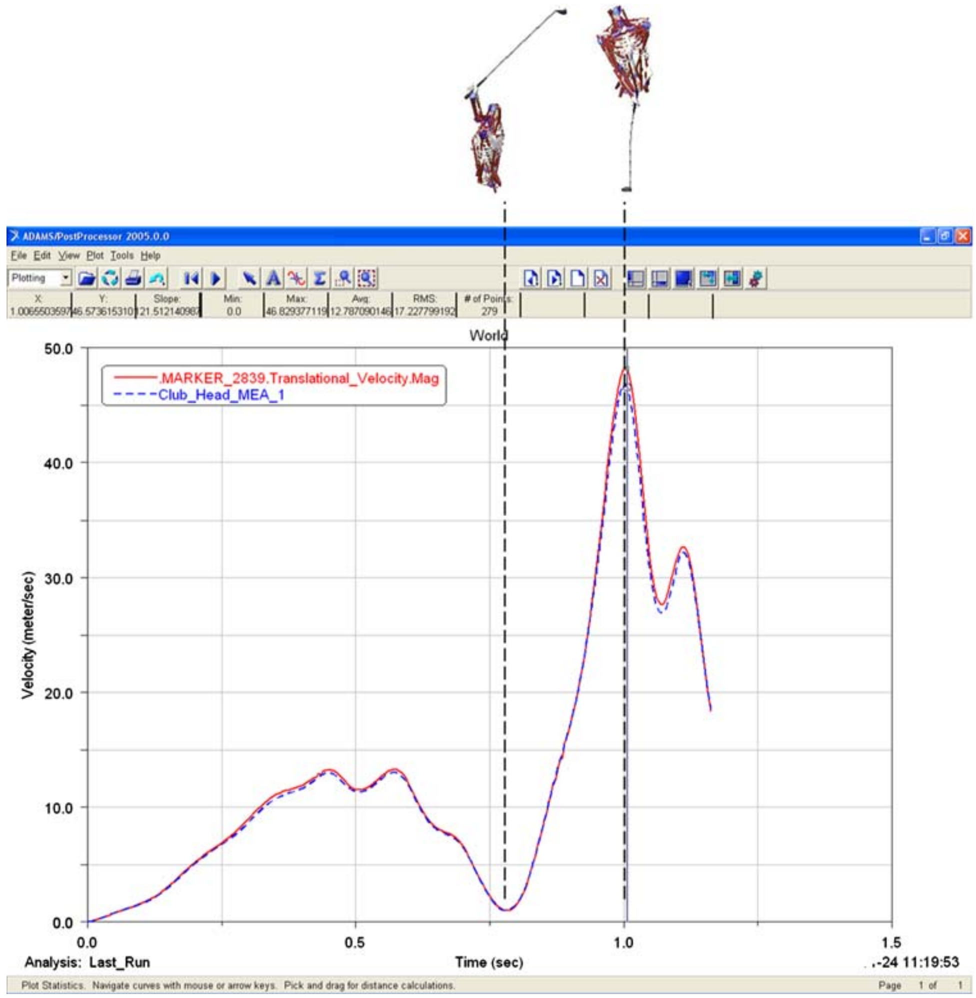


Fig. 4 Representative driver model clubhead velocity against normalised experimental driver clubhead velocity

Table 4 Validation markers/model anatomical landmark correlation

Marker	Pearson's 'r'	RMS difference (°)
R acromion	0.997*	0.06
L acromion	0.997*	0.06
L5	0.966*	0.11
Navel	0.990*	0.05
R greater trochanter	0.995*	0.05
L greater trochanter	0.987*	0.08
R posterior thigh	0.969*	0.06
L posterior thigh	0.929*	0.05
R inferior patella	0.991*	0.05
L inferior patella	0.991*	0.03
R medial malleolus	0.996*	0.02
L medial malleolus	0.990*	0.03

R right, *L* left

* $P < 0.001$ (two-tailed)

Table 5 Comparison of left hand third finger metacarpal joint peak grip force during the swing between a driver model predicted results and previously reported experimental research

Source	Grip force (N)
Model	13.2
Nikonovas et al. [16]	8–17
Budney [5]	13–23

Table 8 compares the total and component durations of backswing and downswing for the 7 iron and driver simulations. It is shown that timing remains approximately the same no matter which club is used.

Table 9 details the relationship between the driver and 7 iron swing for KE output for the four body segments studied. An overall Pearson's correlation of $r = 0.95$ and a root mean square deviation (RMSD) of just 2.04 J was found.

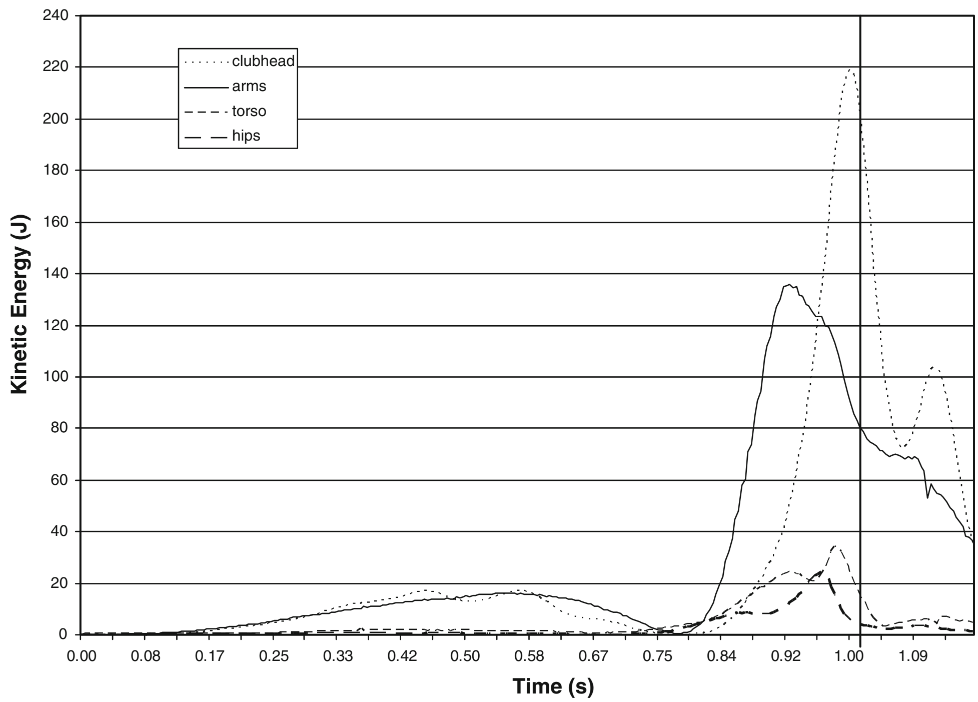


Fig. 5 Kinetic energy for segments throughout the golf swing for a single player during one simulation using a driver. Ball impact is highlighted at time = 1.019 s

Table 6 Absolute and relative timing of peak kinetic energy (KE) for a simulated driver golf swing

Segment	Peak KE (J)*	Time of peak KE (s)	Relative timing to impact[a]
Club	219.0	1.00	−0.019 (98.1%)
Arms	136.0	0.922	−0.097 (90.5%)
Torso	33.9	0.982	−0.037 (96.4%)
Hips	24.4	0.965	−0.054 (94.7%)

* $F = 58.1$, $P < 0.0001$

[a] Impact time = 1.019 s

4 Discussion

Using this novel method of modelling analysis similar results to those found by Anderson et al. [2] have been presented. Peak magnitudes of KE increased sequentially from proximal to distal segments during the golf swing simulations for both the driver and 7 iron. This supports the principle of the summation of speed. Longer club length resulted in a simulated clubhead C.O.M. linear velocity of 48.3 m s^{-1} for the driver compared to 33.8 m s^{-1} for the 7 iron. Variability, or SD of multiple simulations was negligible. Importantly, whilst KE peak magnitude was greater for the driving club segment (219.0 J) than the 7 iron club segment (174.0 J), as would be expected due to longer club length therefore higher distal end linear velocity, torso and arms body segments did not vary significantly when different clubs were used (driver arms 136.0 J, driver torso 33.9 J; 7 iron arms 141.0 J, 7 iron torso 33.4 J). This indicates that the iron swing, at least for the representative golfer studied here, was as forceful as the driver swing. This supports work by previous research [6, 8] indicating that the golf swing should be performed in the same way no matter which club is used. It could be that for highly skilled golfers for approach shots to the green using irons, the same level of consideration is given to control as is for driver shots, that is very high levels of skill do not require a reduction in iron play swing power. However, it should be noted that in this experimental protocol, the laboratory target was only some 4.5 m away from the tee, not representative of either drive or approach shots on the fairway. Further investigation could be carried out on low-medium[2] handicap golfers to ascertain whether lower skilled golfers alter swing power.

The present study does not support the principle of sequentially increased KE from proximal to distal segments [7]. Also, neither does it support the principle of optimum coordination of partial *momenta* [18] which describes simultaneous peak angular velocity for linked segments. Rather, a subject-specific pattern firstly of peak arms KE, followed by peak hips, torso and club KE was exhibited. Adlington [1] stated that “There is no one swing for everybody, but everybody must have one swing” (p. 10), supporting the use of single-subject, or subject-specific investigation. In recent years biomechanical studies have been carried out on single subjects [3, 4, 14, 20]. It has been reported, both in experimental and theoretical

[2] Low: 0–5 handicap (category 1); medium: 6–12, 13–20 handicap (categories 2 and 3) [9].

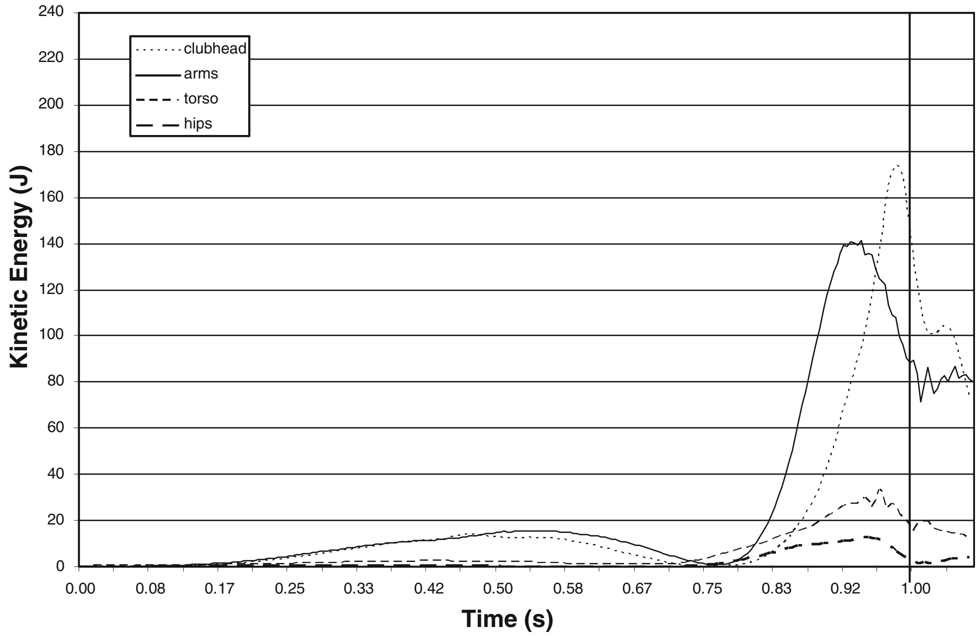


Fig. 6 Kinetic energy for segments throughout the golf swing for a single player during one simulation using a 7 iron. Ball impact is highlighted at time = 0.9958 s

Table 7 Absolute and relative timing of peak KE for a simulated 7 iron golf swing

Segment	Peak KE (J)*	Time of peak KE (s)	Relative timing to impact[a]
Club	174 .0	0.979	−0.0168 (98.3%)
Arms	141.0	0.938	−0.0578 (94.2%)
Torso	33.4	0.958	−0.0378 (96.2%)
Hips	12.8	0.946	−0.0498 (95.0%)

* $F = 43.7$, $P < 0.0001$

[a] Impact time 0.9958 s

Table 8 Total and component swing times

Club	Total duration (s)	BS duration (s)	DS duration (s)
Driver	1.0190	0.7850	0.2340
7 Iron	0.9958	0.7500	0.2458

BS backswing, *DS* downswing

Table 9 Pearson's correlation and RMSD between a 7 iron and a driver for segment KE during a simulated golf swing

Statistic	Club	Arms	Torso	Hips
Pearson's r	0.949*	0.997*	0.933*	0.917*
RMSD (J)	0.409	5.925	0.818	1.021

* Significant at the 0.01 level

modelling journal papers, that it is unlikely that any two golfers will have an identical swing, and even that an individual golfer is unlikely to produce two identical swings in terms of kinematics. Also, naturally, intra-subject trial data will usually correlate better than inter-subject data. The huge number of degrees-of-freedom associated with whole body movements, and the larger number of motor control units and muscles involved in multi-joint movements mean that the method by which a golfer moves the clubhead from the address position to make appropriate impact with the ball will differ in three-dimensional space. Results seem to indicate a subjective optimal coordination of sequencing, reinforcing the benefit of single-subject analysis in the study of the golf swing.

Finally, similar to Anderson et al. [2] a decrease in proximal KE seemed to occur when the club segment KE reached a maximum peak. It should be expected that if energy were being transferred sequentially along body segments (optimum coordination, Putnam [18] then distal links should experience an increase in KE at the expense of their proximal link.

5 Conclusion

Results seems to indicate a subjective optimal coordination of sequencing, demonstrating the summation of speed principle, but not fully explaining or adhering to this principle or the optimal control of partial *momenta*. Through comparison of KE for the driver and 7 iron swing using this novel method of computer simulation, findings indicate that for a highly skilled golfer, an iron may be swung with as much force as a driver.

Acknowledgments We would like to express our thanks to the R&A Rules Limited for their valuable contributions to the study.

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