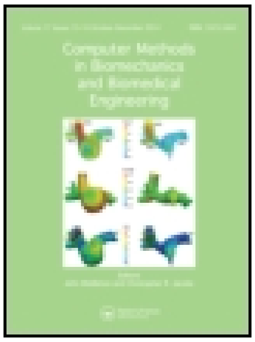

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A methodological approach for the biomechanical cause analysis of golf-related lumbar spine injuries

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A new methodological approach employing mechanical work (MW) determination and relative portion of its elemental analysis was applied to investigate the biomechanical causes of golf-related lumbar spine injuries. Kinematic and kinetic parameters at the lumbar and lower limb joints were measured during downswing in 18 golfers. The MW at the lumbar joint (LJ) was smaller than at the right hip but larger than the MWs at other joints. The contribution of joint angular velocity (JAV) to MW was much greater than that of net muscle moment (NMM) at the LJ, whereas the contribution of NMM to MW was greater rather than or similar to that of JAV at other joints. Thus, the contribution of JAV to MW is likely more critical in terms of the probability of golf-related injury than that of NMM. The MW-based golf-related injury index (MWGII), proposed as the ratio of the contribution of JAV to MW to that of NMM, at the LJ (1.55) was significantly greater than those at other joints (< 1.05). This generally corresponds to the most frequent occurrence of golf-related injuries around the lumbar spine. Therefore, both MW and MWGII should be considered when investigating the biomechanical causes of lumbar spine injuries.

Keywords: mechanical work; net-muscle moment; joint angular velocity; lumbar spine injury; mechanical work-based golf-related injury index

1. Introduction

As golf becomes more popular regardless of age, gender or athletic ability, the number of golf-related injuries has also risen (Theriault and Lachance 1998; Parziale and Mallon 2006). A golf swing is a complex motion, which requires power, accuracy and consistency (Milburn 1982). Especially, the modern golf swing, pursuing power and distance, demands greater axial twisting motion of the lumbar (Hosea et al. 1994; Gluck et al. 2008). This stressful swing mechanism is a potential cause of low back pain (LBP), which is one of the most common golf-related symptoms (Stover et al. 1976; McCarroll and Gioe 1982). Golfers with and without LBP exhibited differences in musculoskeletal motion during their golf swing (Horton et al. 2001; Suter and Lindsay 2001; Grimshaw and Burden 2002; Lindsay et al. 2002; Vad et al. 2004).

The greater axial twisting motion resulting from the modern powerful swing can produce a greater angular velocity of the lumbar spine (Zheng et al. 2008; i.e. ~ 600–$800°$/s determined). This is consistent with the fact that an increased *x*-factor, that is the difference in the angle formed by the large shoulder rotation versus the restrictive hip turn, leads to a very high angular displacement of the lumbar spine (Lindsay et al. 2002; Parziale and Mallon 2006). Such an excessive angular velocity, when it surpasses a threshold in terms of joint mobility, has the potential to induce injuries to the musculoskeletal structure of the lumbar spine (Zheng et al. 2008; Gulgin et al. 2009). In general, the golf swing generates tremendous amounts of torque load on the lumbar spine (Stover et al. 1976; Hosea and Gatt 1996), resulting in high stress on the muscles surrounding the lumbar spine, which can cause LBP in susceptible golfers (Parziale 2002; Brandon and Pearce 2009). Repeated stressful loads on the lumbar spine greater than its biomechanical tolerance are recognised as major risk factors for lower back injuries (Pink et al. 1993; Cabri et al. 2009). Thus, both angular velocity and torque, which are directly associated with lumbar spine injury, should be considered in analyses of the effects of golf-swing mechanics on the structures and injuries of the lumbar spine.

An integrated term taking account of both the angular velocity and torque will represent the biomechanical work done at a joint during the golf-swing motion. When analysing human movement, mechanical work (MW) is generally defined as an integration of a vector product of joint angular velocity (JAV) and net muscle moment (NMM) (Winter 1979). The JAV is calculated by numerical differentiation of the time-based angle data of a joint, and is related to the dynamics of muscle activation and force generation (Granata and Abel 2000). The NMM is the sum of the torque produced by all muscles crossing a joint, and represents the motor patterns used to produce human movement (Winter 1983). MW determined based on this definition is very accurate because the inclusion of

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kinetic data (such as NMM) compensates for the inaccuracy caused by kinematic data (such as JAV) (Arampatzis et al. 2000). Since the MW has often been utilised to analyse the efficiencies and injuries associated with a variety of human movements (Winter 1979; Hubley and Wells 1983; Kaneko 1990; Thys et al. 1996; Arampatzis et al. 2000; Kim et al. 2006), it would be a good primary parameter for a biomechanical understanding of the cause-and-effect relationships of golf-swing mechanics (Nesbit and Serrano 2005) because of its inclusion of both kinematic and kinetic factors at three-dimensional (3D) axes. The MW value determined at a joint is related to muscle fatigue and injury possibility (Anderson and Pandy 2001; Kim et al. 2006); however, it cannot be directly compared with its biomechanical injury tolerance. Because each MW element has a direct impact on injury potential, the contribution of either JAV or NMM to MW must be assessed to elucidate which element has a greater impact on an inherent but variable joint injury threshold. When interpreting the meaning of the MW value, evaluating the relative contributions of both elements to MW is important because such interactions can be considered in a portion analysis.

Therefore, in this study, the MWs at the lumbar and lower limb joints during the downswing were first analysed, and then the relative contributions of each element to the MW were evaluated to investigate the possible biomechanical causes of golf-related injuries to the lumbar spine. We focused on a biomechanical analysis of the downswing motion because golf-related injuries occur most frequently during this phase of the swing (Adlington 1996; Burden et al. 1998). Furthermore, we propose a MW-based golf-related injury index (MWGII) determined using this novel methodological approach that employed MW determination and analysis of the relative contributions of its constituent elements.

2. Methods

2.1 Participants

Nine professional (eight males and one female) and nine amateur (seven males and two females) golfers participated in this study (mean age: 36.5 ± 9.6 years; height: 1.73 ± 0.06 m; body weight: 75.5 ± 7.8 kg; handicap index: 5.7 ± 6.7). All right-handed participants ($n = 18$) had no significant history of injury. Informed consent was obtained from all participants in accordance with local regulations.

2.2 Instrumental system and experimental procedures

A 3D motion analysis system including six infrared MCam2 cameras (VICON 460 system; Oxford Metrics Ltd, Oxford, UK) and two force plates (AMTI, Watertown, MA, USA) was used (Figure 1). Twenty-four reflective markers (25 mm diameter) were attached to the body (20 markers) and golf club (four markers) (Figure 2). Body measurement information (ankle thickness, knee thickness and leg length) was converted to the calculated joint centres of the lumbar spine and lower limbs using the Newton–Euler equation human modelling method (Choi et al. 2005, 2006). 3D motion was captured at 120 Hz, and force data from the system were synchronised to time. Kinematic and kinetic parameters of golf-swing motion were calculated using a VICON 460 system. After participants had fully warmed up by performing several practice swings using their own drivers, 10 full-swing motions per participant were captured for raw data acquisition under identical conditions.

2.3 Data reduction

The total power ($|P(t)|$) of a joint is defined by the absolute value of the dot product of vectors of NMM and JAV at

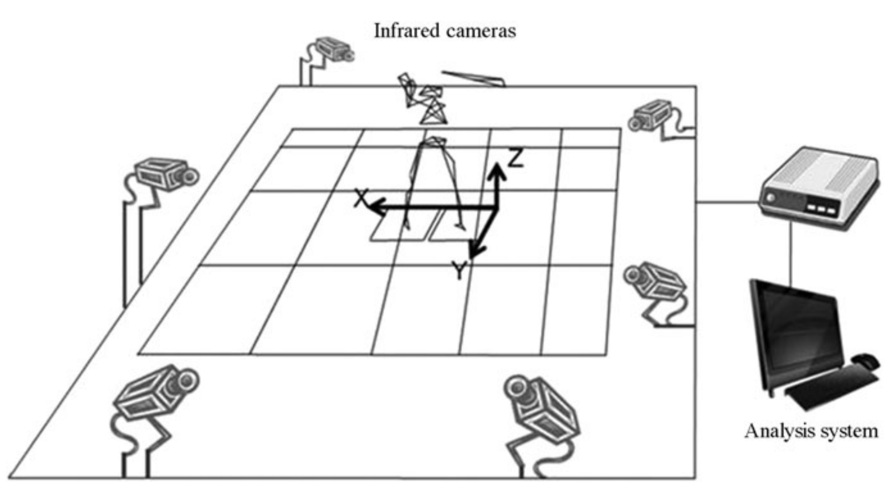

Figure 1. Instrumental system and experimental set-up.

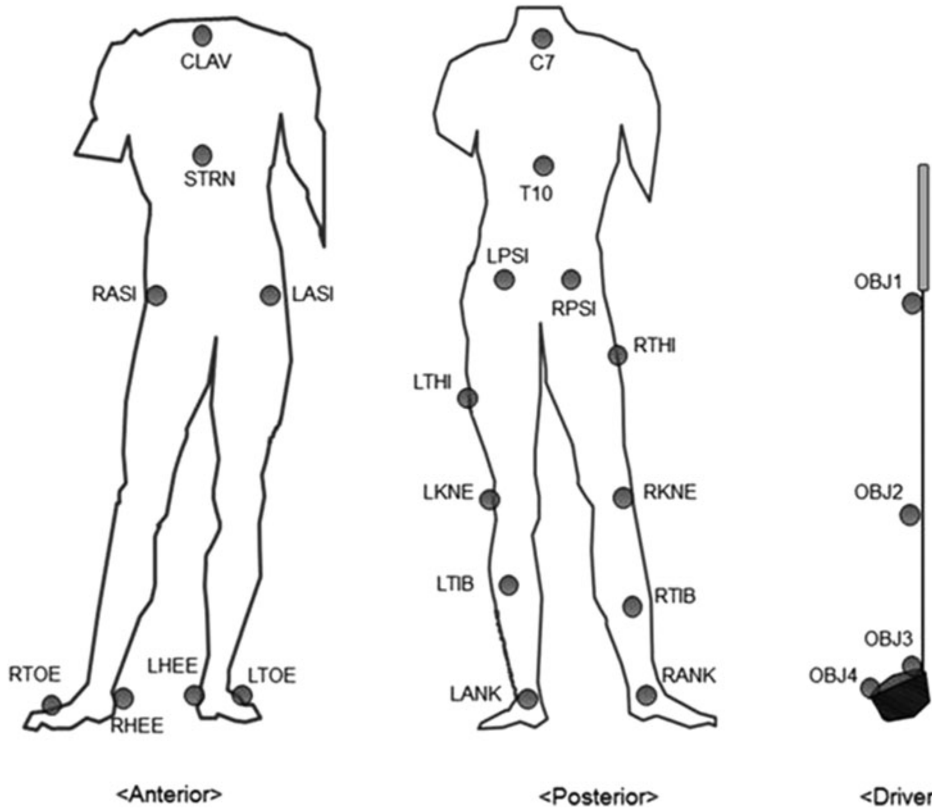

Figure 2. Retro-reflective marker configuration utilised for motion analysis of golf swings (CLAV: clavicle; STRN: sternum; C7: cervical vertebra 7th; T10: thoracic vertebra 10th; LASI: left anterior superior iliac; RASI: right anterior superior iliac; LPSI: left posterior superior iliac; RPSI: right posterior superior iliac; LTHI: left thigh; RTHI: right thigh; LKNE: left knee; RKNE: right knee; LTIB: left tibia; RTIB: right tibia; LANK: left ankle; RANK: right ankle; LHEE: left heel; RHEE: right heel; LTOE: left toe; RTOE: right toe; OBJ1–4: objects 1–4).

time (t). The NMM and JAV were generated from the raw data by the motion analysis system. As described in Section 1, MW is an integration of the total power over time as shown in Equation (1) (Arampatzis et al. 2000; Kim et al. 2006; Ren et al. 2007):

$$\mathrm{MW} = \int_{t_1}^{t_n} |P(t)|\,\mathrm{d}t = \int_{t_1}^{t_n} \left|[\overline{\tau(t)}][\overline{\dot{q}(t)}]^{\mathrm{T}}\right|\mathrm{d}t, \tag{1}$$

where t_1 and t_n are the start and end times of the downswing, respectively, $\overline{\tau(t)}$ and $\overline{\dot{q}(t)}$ represent NMM (N m/kg) normalised to body weight and JAV (rad/s), respectively. Since the true MW value could not be realistically obtained under this experimental system, approximate MW values were calculated by multiplying the sum of the total power during the downswing by the motion capturing time interval ($\Delta t = 120$ Hz), as shown in equation.

$$\mathrm{MW} \cong \Delta t \cdot \sum_{t=t_1}^{t_n} |\tau_x(t)\dot{q}_x(t) + \tau_y(t)\dot{q}_y(t) + \tau_z(t)\dot{q}_z(t)|, \tag{2}$$

where the subscripts, x, y and z, represent the individual 3D axes.

2.4 *Statistical analysis*

Analysis of variance (ANOVA) was used to detect overall significant differences in MW and MWGII values between joints and in contribution of between JAV and NMM to MW for all participants tested ($n = 18$). The significance level for all ANOVAs was set at 5%. Independent t-test was used for *post hoc* analysis at a significant level of 0.05. The statistical analysis system (SAS Institute, Inc., Cary, NC, USA) was used for data analysis.

3. Results and discussion

3.1 *Impact of MW on golf-related injuries*

Average MW and standard deviation values at the lumbar and lower limb joints during the downswing for all participants are shown in Figure 3. The average MW values at the (RH) were largest, those at the right hip lumbar joint (LJ) and left hip (LH) were large and those at the remaining joints were relatively small. The average magnitude order of the MW at joints, which was identical in all participants, was RH > LJ > LH > knee > ankle. The lack of variation in the average MW magnitude order

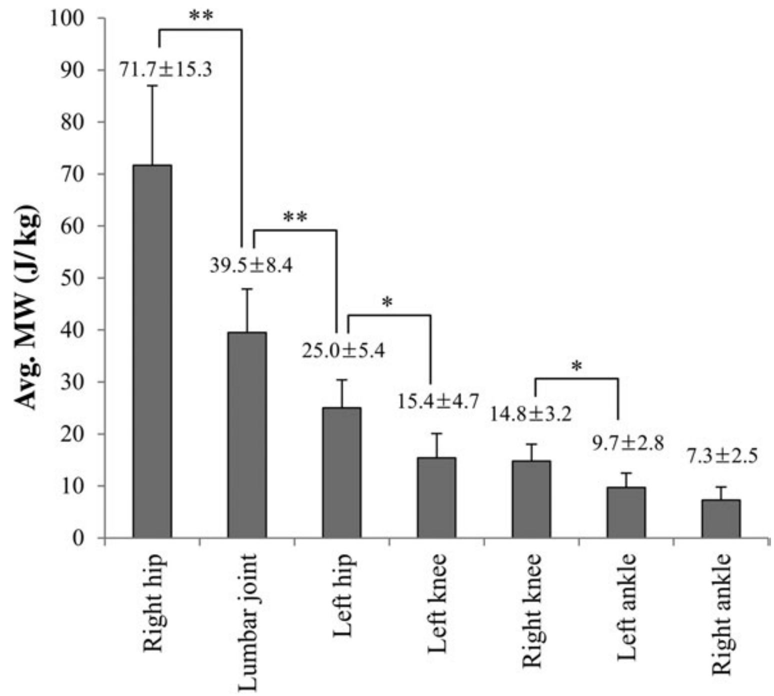

Figure 3. Average MW at each joint during the downswing in all participants ($n = 18$). Statistical significance was evaluated for differences in MW value between joints: $^{*}p < 0.05$ and $^{**}p < 0.001$.

was verified by ANOVA ($p < 0.05$). Also, the magnitude order shows a tendency towards larger MWs at joints more distal from the ground. This is fairly consistent with the so-called principles of summation, i.e. that the golf swing should generate faster movement of the more distal segments (Bunn 1972; Milburn 1982).

Previous studies on golf-related injury distributions represented that LJ injuries were much more probable than other lower joint injuries during golf swing (Table 1) (McCarroll and Gioe 1982; Gosheger et al. 2003; Fradkin et al. 2007). As shown in Figure 3, the average MW value at the LJ was larger than at other joints (with the exception of the RH). The comparison of this result with the previous findings indicates that a larger MW is associated with a higher likelihood of golf-related joint injury. By the way, golf-related injuries generally occur far less frequently at the RH than at the LJ, while the MW values were at the RH larger than those at the LJ (Table 1 and Figure 3). This observation suggests no strictly proportional correlation of absolute MW magnitude to injury probability, which means that the MW magnitude is unlikely to be an unconditional indicator of injury probability. Therefore, other factors and aspects besides absolute MW magnitude itself are to be considered so that the biomechanical causes of golf-related LJ injuries can be more clearly analysed. Nevertheless, it cannot be denied that the biomechanical analysis of golf-related injuries by MW determination may provide a tendency and a macroscopic estimate of injury probability to the corresponding joint.

Table 1. Comparison of the average MW values at the lumbar and lower limb joints during the downswing ($n = 18$) with previously reported golf-related injury distributions (McCarroll and Gioe 1982; Gosheger et al. 2003; Fradkin et al. 2007).

		Injury distribution (%)		
Site	Average MW (J/kg)	McCarroll and Gioe	Gosheger et al.	Fradkin et al.
LJ	39.5	23.7	16.3	36.0
RH	71.7	1.0	2.8	1.8
LH	25.0			
Right knee	14.8	6.6	3.6	7.2
Left knee	15.4			
Right ankle	7.3	2.0	4.6	5.4
Left ankle	9.7			

3.2 Contribution of MW elements

Excess JAV or NMM can result in golf-related injury to joints when it exceeds the biomechanical tolerance (Pink et al. 1993; Cabri et al. 2009), as stated above. The rise in MW is the result of an increase in JAV and/or NMM. As such, the relative contribution of either JAV or NMM to MW must be assessed to understand the biomechanical meaning of MW with respect to the likelihood of joint injury and to determine which MW element has a greater impact on the inherent but variable joint injury threshold. Based on the MW concept, the contribution of either JAV or NMM to MW can be defined as its contribution to total power during the downswing over the MW. We propose the following theoretical approach to derive the contribution of the JAV or NMM portion.

Because the units and data ranges of JAV and NMM differ, their raw data are first normalised to be 0 to 1, using the following equation:

$$\hat{q}_i(t) = \frac{\dot{q}_i(t) - \dot{q}_i^{\min}}{\dot{q}_i^{\max} - \dot{q}_i^{\min}} \quad (i = X,\ Y \text{ or } Z \text{ axis}),$$

where $\dot{q}_i(t)$ is the measured JAV value, $\hat{q}_i(t)$ is JAV normalised at time (t) and $\dot{q}_i^{\max}$ and $\dot{q}_i^{\min}$ are the maximum and minimum JAV values, respectively. The normalised NMM, $\hat{\tau}_i(t)$, is derived in the same way. A theoretical term of total power, normalised total power $(\hat{P}(t))$, is calculated by putting normalised JAV and NMM values into Equation (3).

$$\hat{P}(t) = \hat{\tau}_x(t)\hat{q}_x(t) + \hat{\tau}_y(t)\hat{q}_y(t) + \hat{\tau}_z(t)\hat{q}_z(t). \tag{3}$$

The JAV portion contribution to the x-axis component of the normalised total power at time (t) is theoretically defined as

$$\left(\hat{\tau}_x(t)\hat{q}_x(t)\right)\left\{\frac{\hat{q}_x(t)}{\hat{\tau}_x(t) + \hat{q}_x(t)}\right\},$$

where $\hat{\tau}_x(t)\hat{q}_x(t)$ and $(\hat{q}_x(t))/(\hat{\tau}_x(t) + \hat{q}_x(t))$ represent the x-axis component of the normalised total power and the occupancy ratio of JAV to the sum of JAV and NMM, respectively. Therefore, the JAV contribution to the normalised total power at a time is

$$\left(\hat{\tau}_x(t)\hat{q}_x(t)\right)\left\{\frac{\hat{q}_x(t)}{\hat{\tau}_x(t) + \hat{q}_x(t)}\right\} + \left(\hat{\tau}_y(t)\hat{q}_y(t)\right)\left\{\frac{\hat{q}_y(t)}{\hat{\tau}_y(t) + \hat{q}_y(t)}\right\}$$
$$+ \left(\hat{\tau}_z(t)\hat{q}_z(t)\right)\left\{\frac{\hat{q}_z(t)}{\hat{\tau}_z(t) + \hat{q}_z(t)}\right\},$$

and the integration of this expression from t_1 to t_n represents the JAV portion contribution to normalised total power during the downswing. Finally, the contribution of JAV to MW $[C_{\hat{q}}(\%)]$ can be determined by dividing the JAV contribution to the normalised total power during the downswing (numerator) by the integration of the normalised total power over downswing time (denominator), as in Equation (4).

$$C_{\hat{q}}(\%) = \frac{\int_{t_1}^{t_n}\left\{\hat{\tau}_x(t)\hat{q}_x(t)\frac{q_x(t)}{\hat{\tau}_x(t)+\hat{q}_x(t)} + \hat{\tau}_y(t)\hat{q}_y(t)\frac{\hat{q}_y(t)}{\hat{\tau}_y(t)+\hat{q}_y(t)} + \hat{\tau}_z(t)\hat{q}_z(t)\frac{\hat{q}_z(t)}{\hat{\tau}_z(t)+\hat{q}_z(t)}\right\}}{\int_{t_1}^{t_n}\{\hat{\tau}_x(t)\hat{q}_x(t) + \hat{\tau}_y(t)\hat{q}_y(t) + \hat{\tau}_z(t)\hat{q}_z(t)\}\mathrm{d}t} \times 100. \tag{4}$$

The approximate $C_{\hat{q}}(\%)$ can also be calculated using Equation (5).

$$C_{\hat{q}}(\%) \cong \frac{\sum_{t_1}^{t_n}\left\{\hat{\tau}_x(t)\hat{q}_x(t)\frac{\hat{q}_x(t)}{\hat{\tau}_x(t)+\hat{q}_x(t)} + \hat{\tau}_y(t)\hat{q}_y(t)\frac{\hat{q}_y(t)}{\hat{\tau}_y(t)+\hat{q}_y(t)} + \hat{\tau}_z(t)\hat{q}_z(t)\frac{\hat{q}_z(t)}{\hat{\tau}_z(t)+\hat{q}_z(t)}\right\}}{\sum_{t_1}^{t_n}\{\hat{\tau}_x(t)\hat{q}_x(t) + \hat{\tau}_y(t)\hat{q}_y(t) + \hat{\tau}_z(t)\hat{q}_z(t)\}} \times 100. \tag{5}$$

The same approach can be applied to the determination of the contribution of NMM to MW $[C_{\hat{\tau}}(\%)]$, as shown in Equation (6).

$$\begin{aligned} C_{\hat{\tau}}(\%) &\cong \frac{\sum_{t_1}^{t_n}\left\{\hat{\tau}_x(t)\hat{q}_x(t)\frac{\hat{\tau}_x(t)}{\hat{\tau}_x(t)+\hat{q}_x(t)} + \hat{\tau}_y(t)\hat{q}_y(t)\frac{\hat{\tau}_y(t)}{\hat{\tau}_y(t)+\hat{q}_y(t)} + \hat{\tau}_z(t)\hat{q}_z(t)\frac{\hat{\tau}_z(t)}{\hat{\tau}_z(t)+\hat{q}_z(t)}\right\}}{\sum_{t_1}^{t_n}\{\hat{\tau}_x(t)\hat{q}_x(t) + \hat{\tau}_y(t)\hat{q}_y(t) + \hat{\tau}_z(t)\hat{q}_z(t)\}} \times 100 \\ &= 100 - C_{\hat{q}}(\%). \end{aligned} \tag{6}$$

The $C_{\hat{q}}(\%)$ and $C_{\hat{\tau}}(\%)$ results calculated using Equations (5) and (6) are shown in Table 2. The average $C_{\hat{q}}(\%)$ (60.8%) was much greater than the average $C_{\hat{\tau}}(\%)$ (39.2%) at the LJ, where a moderately high MW was observed and golf-related injuries occurred most frequently. Both the average $C_{\hat{q}}(\%)$ and $C_{\hat{\tau}}(\%)$ were similar within their range of error at the LH, left/right knees and left/right ankles, where relatively low MWs were detected and golf-related injury probabilities were low. In contrast, the average $C_{\hat{q}}(\%)$ (43.8%) was smaller than the average $C_{\hat{\tau}}(\%)$ (56.2%) at the RH where golf-related injuries occurred only rarely, despite a very high MW. These results suggest that golf-related injuries may have occurred more often at joints with larger MW and greater $C_{\hat{q}}(\%)$. Also, golf-related injuries may occur less often at joints with a $C_{\hat{q}}(\%)$ smaller than $C_{\hat{\tau}}(\%)$ or similar $C_{\hat{q}}(\%)$ and $C_{\hat{\tau}}(\%)$ values, regardless of the comparative MW magnitude. In summary, the contribution of JAV to MW obtained from the MW-based calculation (Equation (5)) is very likely more critical to the probability of golf-related injuries than that of NMM. Therefore, not only MW

Table 2. Contribution values (%) of JAV and NMM to MW at the lumbar and lower limb joints during downswing in all participants tested ($n = 18$). Statistical p-values apply to differences in contribution of between JAV and NMM to MW.

	Contribution (%)		
Site	JAV (Avg. ± SD)	NMM (Avg. ± SD)	p-Value
LJ	60.8 ± 8.7	39.2 ± 8.7	0.000
RH	43.8 ± 3.7	56.2 ± 3.7	0.000
LH	50.5 ± 1.9	49.5 ± 1.9	0.120
Right knee	49.0 ± 2.0	51.0 ± 2.0	0.004
Left knee	51.2 ± 1.7	48.8 ± 1.7	0.000
Right ankle	51.2 ± 2.1	48.8 ± 2.1	0.001
Left ankle	51.0 ± 1.8	49.0 ± 1.8	0.002

magnitude but also JAV contribution to the MW must be determined when evaluating the probability of golf-related joint injury.

3.3 JAV contribution to MW

The relative contribution of JAV to MW generated at a joint during the downswing was analysed to determine whether it is an indicator of the probability of golf-related injury. As shown in Equations (5) and (6), the greater the $C_{\hat{q}}(\%)$ is, the smaller the $C_{\hat{\tau}}(\%)$ is. In this study, the relative contribution of JAV to MW was defined as the ratio of the contribution of JAV to MW to that of NMM, which was named the 'MW-based golf-related injury index' (MWGII):

$$\mathrm{MWGII} = \frac{C_{\hat{q}}(\%)}{C_{\hat{\tau}}(\%)} = \frac{C_{\hat{q}}(\%)}{100(\%) - C_{\hat{q}}(\%)}.$$

The MWGII values determined at the lumbar and lower limb joints are shown in Table 3. If $C_{\hat{q}}(\%)$ is not different from $C_{\hat{\tau}}(\%)$, MWGII is equal or near to unity. As presented in Figure 4, the average MWGII values at the LH, left/right knees and left/right ankles were near unity because their JAV and NMM contributions to MW were similar within their ranges of error. The average MWGII at the LJ (1.55) was considerably greater than unity because its $C_{\hat{q}}(\%)$ was much greater than $C_{\hat{\tau}}(\%)$. The average MWGII at the RH (0.78) was less than unity because its $C_{\hat{q}}(\%)$ was much smaller than $C_{\hat{\tau}}(\%)$. Thus, if the MWGII value is far greater than unity, the probability of a golf-related injury is high, whereas if the MWGII value is near or less than unity, the potential of a golf-related injury is low.

The higher frequency of golf-related injuries at the LJ can be more clearly understood when both MW magnitude and MWGII value are considered: its relatively large MW and MWGII are far greater than unity. Such a large MW and the greater contribution of JAV to MW are the result of a large rotational and excessive twisting swing motion at the LJ. In addition, the lower frequency of golf-related injuries at other joints may be due to the smaller MWGII

Table 3. MWGII values at the lumbar and lower limb joints during the downswing in all participants tested ($n = 18$).

Site	Avg. ± SD	Range Min.	Range Max.
LJ	1.55 ± 0.16	1.40	1.90
RH	0.78 ± 0.06	0.64	0.90
LH	1.02 ± 0.07	0.92	1.15
Right knee	0.96 ± 0.09	0.79	1.13
Left knee	1.05 ± 0.10	0.85	1.19
Right ankle	1.05 ± 0.07	0.89	1.14
Left ankle	1.04 ± 0.07	0.90	1.11

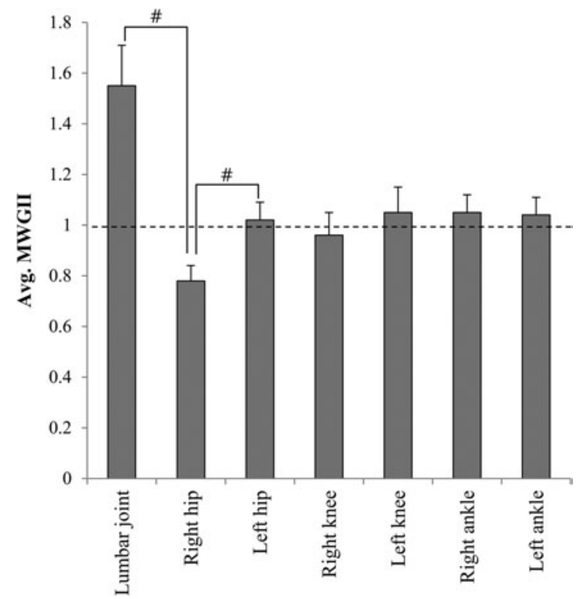

Figure 4. Average MWGII at each joint for all participants ($n = 18$). Statistical significance was evaluated for differences in MWGII value between joints: $^{\#}p < 0.001$.

($\leq \sim 1$), irrespective of MW magnitude. No excessive twisting swing motion is generally observed at the RH or remaining joints, whereas the rotation velocity is higher at the RH compared to other joints. Thus, MWGII is a more accurate estimate of the probability of golf-related injuries at each joint than MW magnitude alone, and seems to be a casual indicator. The probability of golf-related injuries cannot be precisely determined using MW magnitude alone, even though it provides a rough estimate. For example, golf-related injuries occur at the RH only rarely due to a MWGII smaller than unity, although its MW magnitude is very high (Figure 3). That is, a larger MW does not necessarily indicate a higher probability of injury.

3.4 A new approach for analysing the biomechanical causes of golf-related lumbar spine injuries

This study was focused on investigating biomechanical aspects why golf-related LJ injuries are, in general, most frequent to golfers, not a particular golfer. We propose that the reason can be explained based on the MW and MWGII. The high JAV and NMM generated at the LJ from the powerful swing used in modern golf generally lead to a large MW. A sufficiently large MW indicates stressful swing mechanics resulting from maximising the power of the golf swing, and the likelihood of lumbar spine injury increases. MW must be determined to estimate injury probability. Since the probability of golf-related LJ injuries cannot be precisely assessed using MW magnitude alone, the MWGII must also be determined. We showed that MWGII, the relative contribution of JAV to MW, was a causal indicator of the probability of golf-related injuries. MWGII is a MW-based biomechanical factor, but is independent of MW magnitude. The probability of golf-related lumbar spine injuries is high if the MWGII is far greater than unity and the MW is relatively large.

In general, golfers with LBP or a history of serious back injury could hardly demonstrate powerful swing mechanics as much as healthy, asymptomatic golfers could do. Previous investigations (Vad et al. 2004; Tsai et al. 2010; Kalra et al. 2012) reported that the rotational angle of motion, angular velocity and strength (moment) at lumbar spine measured during the golf swing of golfers with LBP decreased compared to those with no LBP. These indicate that the decreased MW values at LJ are anticipated in LBP golfers. However, whether the MWGII was unconditionally decreased or increased due to LBP is not known. Instead the MWGII value for each LBP golfer would depend on the sites and severity of injuries around lumbar spine. It is unlikely that swing motions with excessive spinal rotation velocity for powerful swing, which leads to the MWGII value much greater than unity, may be made by golfers with LBP. Likewise, no swing motions with a combination of low MW and high MWGII or very low MWGII at the LJ almost realistically happen to asymptomatic golfers. MWGII estimation along with MW determination would be very useful in analysing the golf swing mechanics of golfers with LBP or lumbar spine injuries, but could not provide direct biomechanical information on why lumbar spine injuries occurred to the same golfers who had never been injured.

In conclusion, this new methodological approach using MW determination and analysis of the relative contribution of its elements can be applied to assess the biomechanical causes of golf-related lumbar spine injury. Our data provide valuable information regarding the prevention of golf-related lumbar spine injuries.

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Notes

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References

Adlington GS. 1996. Proper swing technique and biomechanics of golf. Clin Sport Med. 15(1):9–26.

Anderson FC, Pandy MG. 2001. Static and dynamic optimization solutions for gait are practically equivalent. J Biomech. 34(2):153–161.

Arampatzis A, Knicker A, Metzler V, Brüggemann GP. 2000. Mechanical power in running: a comparison of different approaches. J Biomech. 33(4):457–463.

Brandon B, Pearce PZ. 2009. Training to prevent golf injury. Curr Sport Med Rep. 8(3):142–146.

Bunn J. 1972. Scientific principles of coaching. Englewood Cliffs, NJ: Prentice-Hall.

Burden AM, Grimshaw PN, Wallace ES. 1998. Hip and shoulder rotations during the golf swing in sub-10 handicap players. J Sport Sci. 16(2):165–176.

Cabri J, Sousa JP, Kots M, Barreiros J. 2009. Golf-related injuries: a systematic review. Eur J Sport Sci. 9(6):353–366.

Choi AR, Kim YJ, Rim YH, Kang TG, Min KK, Bae JH. 2006. Development of a spine kinematic model for the clinical estimation of abnormal curvature. Proceedings of the World Congress on Medical Physics and Biomedical Engineering; Seoul, Korea.

Choi AR, Rim YH, Kim YS, Mun JH. 2005. A position based kinematic method for the analysis of human gait. J Mech Sci Technol. 19(10):1919–1931.

Fradkin AJ, Windley TC, Myers JB, Sell TC, Lephart SM. 2007. Describing the epidemiology and associated age, gender and handicap comparisons of golfing injuries. Int J Injury Control Safe Promot. 14(4):264–266.

Gluck GS, Bendo JA, Spivak JM. 2008. The lumbar spine and low back pain in golf: a literature review of swing biomechanics and injury prevention. Spine J. 8(5):778–788.

Gosheger G, Liem D, Ludwig K, Greshake O, Winkelmann W. 2003. Injuries and overuse syndromes in golf. Am J Sport Med. 31(3):438–443.

Granata KP, Abel MF. 2000. Joint angular velocity in spastic gait and the influence of muscle-tendon lengthening. J Bone Joint Surg. 82(2):174–186.

Grimshaw PN, Burden AM. 2002. Case study: reduction of low back pain in a professional golfer. Med Sci Sport Exerc. 32(10):1667–1673.

Gulgin H, Armstrong C, Gribble P. 2009. Hip rotational velocities during the full golf swing. J Sport Sci Med. 8(2):296–299.

Horton JF, Lindsay DM, Macintosh BR. 2001. Abdominal muscle activation of elite male golfers with chronic low back pain. Med Sci Sport Exerc. 33(10):1647–1654.

Hosea TM, Gatt CJ, Jr. 1996. Back pain in golf. Clin Sport Med. 15(1):37–53.

Hosea TM, Gatt CJ, Gertner E. 1994. Biomechanical analysis of the golfers back. In: Stover CN, McCarroll JR, Mallon WJ, editors. Feeling up to par: medicine from tee to green. Philadelphia, PA: Davis FA.

Hubley CL, Wells RP. 1983. A work-energy approach to determine individual joint contributions to vertical jump performance. Eur J Appl Physiol. 50(2):247–254.

Kalra N, Sing J, Neethi M. 2012. Study of trunk movement deficits in golfers with low back pain. Indian J Physiother Occup Ther. 6(3):141–144.

Kaneko M. 1990. Mechanics and energetics in running with special reference to efficiency. J Biomech. 23(1):57–63.

Kim JH, Abdel-Malek K, Yang J, Marler RT. 2006. Prediction and analysis of human motion dynamics performing various tasks. Int J Hum Factors Model Simul. 1(1):69–94.

Lindsay DM, Horton JF, Paley RD. 2002. Trunk motion of male professional golfers using two different golf clubs. J Appl Biomech. 18(4):366–373.

McCarroll JR, Gioe TJ. 1982. Professional golfers and the price they pay. Phys Sports Med. 10(7):64–70.

Milburn PD. 1982. Summation of segmental velocities in the golf swing. Med Sci Sport Exerc. 14(1):60–64.
Nesbit SM, Serrano M. 2005. Work and power analysis of the golf swing. J Sport Sci Med. 4(4):520–533.
Parziale JR. 2002. Healthy swing: a golf rehabilitation model. Am J Phys Med Rehab. 81(7):498–501.
Parziale JR, Mallon WJ. 2006. Golf injuries and rehabilitation. Phys Med Rehbil Clin N Am. 17(3):589–607.
Pink M, Perry J, Jobe FW. 1993. Electromyographic analysis of the trunk in golfers. Am J Sport Med. 21(3):385–388.
Ren L, Jones RK, Howard D. 2007. Predictive modeling of human walking over a complete gait cycle. J Biomech. 40(7):1567–1574.
Stover CN, Wiren G, Topaz SR. 1976. The modern golf swing and stress syndrome. Phys Sports Med. 4(9):43–47.
Suter E, Lindsay D. 2001. Back muscle fatigability is associated with knee extensor inhibition in subjects with low back pain. Spine. 26(16):361–366.
Theriault G, Lachance P. 1998. Golf injuries—an overview. Sports Med. 26(1):42–57.
Thys H, Willems PA, Saels P. 1996. Energy cost, mechanical work and muscular efficiency in swing-through gait with elbow crutches. J Biomech. 29(11):1473–1482.
Tsai YS, Sell TC, Smoliga JM, Myers JB, Learman KE, Lephart SM. 2010. A comparison of physical characteristics and swing mechanics between golfers with and without a history of low back pain. J Orthop Sport Phys. 40(7):430–438.
Vad VB, Bhat AL, Basrai D, Gebeh A, Aspergren DD, Andrews JR. 2004. Low back pain in professional golfers: the role of associated hip and low back range-of-motion deficits. Am J Sport Med. 32(2):494–497.
Winter DA. 1979. Biomechanics of human movement. Ontario: Waterloo.
Winter DA. 1983. Biomechanical motor patterns in normal walking. J Motor Behav. 15(4):302–330.
Zheng N, Barrentine SW, Fleisig GS, Andrews JR. 2008. Kinematic analysis of swing in pro and amateur golfers. Int J Sports Med. 29(6):487–493.