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# Dehydration kinetics of soft contact lenses: The hidden impact of early wear

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ABSTRACT

In contact lens (CL) wear, maintaining adequate CL hydration is essential to prevent ocular dryness and associated discomfort. In this context, dehydration of worn Etafilcon A and Kalifilcon A CLs hydrated by natural tears was monitored using an in-vitro gravimetric method. A kinetic model was applied to derive the water kinetics coefficient, which emerged as a reliable indicator of the on-eye hydration state and water diffusion mechanisms. A significant reduction in water content and a tendency from a polymer-hindered water diffusion toward free diffusion were observed after just five minutes of wear, suggesting a rapid initial adaptation to the ocular environment that alters the material resistance to water evaporation. After two hours of wear, a similar effect was evident in approximately half of the samples investigated, indicating that a new hydration equilibrium may eventually be reached over time.

## 1. Introduction

One of the key factors influencing the performance of soft contact lenses (CLs) is their ability to retain moisture during wear. Moisture loss from a CL can lead to discomfort, dryness, and reduced wearing time [1–7]. As such, the interaction between water and the lens material during wear is critical for ensuring comfort and ocular health [8,9].

To quantitatively assess water diffusion within CL polymers, in-vitro gravimetric methods have been developed [10–12]. These techniques allow for the comparison of dehydration properties across different lens materials using a quantitative parameter known as the water kinetics coefficient ($d$). In a recent study on unworn CLs [11,12], the coefficient $d$ was shown to correlate with the slope of the dehydration rate curve previously reported by González-Méijome et al. [10]. Specifically, $d$ values of unworn CLs hydrated in saline solution were found to be close to 1 for five different CL materials: the hydrogels Etafilcon A and Omafilcon A, and the silicone-hydrogels Delefilcon A, Senofilcon A, and Kalifilcon A [11].

The interpretation of this value is grounded in a theoretical model previously used in the literature to describe the water sorption behaviour of initially-dehydrated polymeric materials [13,14]. According to this model, the fractional water uptake at time $t$, expressed as $M_t/M_\infty$ (where $M_t$ is the water mass at time $t$ and $M_\infty$ is the total water mass at saturation), can be described as a function of time by the equation $t^d$, where $d$ is the water kinetics coefficient. The ratio $M_t/M_\infty$ during water uptake by the polymer matrix reflects the interplay between water diffusion within the polymer network and the swelling behaviour of the matrix. Frisch [13] and Yang et al. [14] described different cases:

(i) $d = 0.5$ corresponds to Fickian water diffusion, where water freely diffuses and matrix swelling occurs almost instantaneously during hydration.
(ii) $d = 1$ describes a scenario where the matrix relaxation (swelling) is slower than water diffusion, thus limiting hydration due to polymer resistance.

The present investigation focuses not on hydration, but rather on the reverse process, examining the dehydration of an initially hydrated CL. Nonetheless, the kinetics is expected to be governed by the same two factors, i.e., water diffusion and matrix response. During dehydration, water begins to evaporate from the surface layer, while the inner layer compensates via diffusion. Simultaneously, the matrix undergoes shrinkage. Therefore, in this context:

i) $d = 0.5$ corresponds to Fickian diffusion, where water diffusion is free and unhindered with slow or negligible matrix shrinkage.
ii) $d = 1$ indicates fast matrix shrinkage that limits water diffusion and hinders evaporation.

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As mentioned above, previous measurements of $d$ values in unworn CLs soaked with saline solution showed values close to 1 for all five tested materials [11] with slight differences: hydrogel CLs and Kalifilcon A CL showed a slightly higher coefficient compared to the two silicone-hydrogel CLs. In principle, higher $d$ values are desirable, as they imply reduced risk of excessive dehydration during wear.

However, these previous studies were conducted on new, unworn CLs hydrated with saline solution [11]. It remains unclear whether similar kinetics are observed when the CLs are hydrated with natural tears rather than saline solution. Moreover, since CLs typically lose part of their interfacial water once placed on the eye [15–18], it is important to assess whether $d$ remains close to 1 after wear and whether it continues to represent a reliable indicator of the hydration state during actual use.

Therefore, this study addresses this gap by measuring the water kinetics coefficients $d$ in CLs immediately after removal from the eye, while still hydrated with tears. The goal is to compare these values with those from unworn, saline-hydrated CLs and determine whether $d$ accurately reflects the actual CL hydration status under realistic wear conditions and offers insight into the mechanisms of water diffusion and lens dehydration.

Previous investigation showed that both Etafilcon A (a hydrogel) and Kalifilcon A (a silicone hydrogel) exhibited relatively high $d$ coefficients [11]. This observation led to selecting these two materials for further exploration of their behaviour during the early phase of lens wear.

## 2. Materials and methods

### 2.1. Materials

All CLs employed in the present study had back vertex power of −0.50 dioptres (D). A description of the two CL materials investigated is provided in Table 1.

### 2.2. In-vitro dehydration kinetics

To investigate the in-vitro dehydration kinetics, each CL was analyzed under controlled environmental conditions using a dynamic vapor sorption analyzer (Aquadyne DVS, Quantachrome Instruments, USA). Measurements were conducted in an enclosed weighing chamber (temperature 36.0 ± 0.2 °C, relative humidity 60.0 ± 0.1 %). The balance pan was adapted by positioning a plastic polydimethylsiloxane spherical holder (radius of curvature of 7.7 mm), on which the CLs were placed. Dehydration profiles as a function of time were obtained by monitoring the CL mass variation over time ($t$) during the dehydration in vitro.

In the first set of experiments, unworn CLs were taken from their packaging blister and rinsed with 5 mL of preservative-free saline solution (Salisin, Schalcon, Italy). The samples were then placed on the microbalance and analyzed under the controlled conditions described above. Measurements on unworn CLs were carried out starting at time $t = 0$ s from the initial mass $M_0$ equal to the estimated mass at the equilibrium water content (EWC, Table 1), denoted as $M_{EWC}$, which was calculated using the following equation:

$$M_{EWC} = \frac{M_{deh}}{1 - EWC} \quad (1)$$

where $M_{deh}$ is the dry mass of the CL, obtained via lyophilization using a freeze dryer (Scanvac, Analytical control De Mori, Italy) and measured using an analytical balance (Entris, Sartorius, Germany). $M_{deh}$ was found to be 15.4 ± 0.1 mg for Etafilcon A and 13.1 ± 0.4 mg for Kalifilcon A. Based on their known EWC, the corresponding $M_{EWC}$ were calculated by Eq. (1) as 36.7 ± 0.2 mg for Etafilcon A and 29.1 ± 0.9 mg for Kalifilcon A.

In the second set of experiments, worn CLs were analyzed. The study was conducted in accordance with the Declaration of Helsinki and approved by the Ethical Committee of the University of Milano-Bicocca (Protocol n. 484, 14/07/2022). Participants were instructed to insert new CLs taken directly from the blister packaging and to wear them for five minutes before carefully removing them next to the measurement station. This ensured that the lenses were immediately transferred on the microbalance plate and the measurement was initiated within a few seconds from removal. In some cases, the initial mass $M_0$ at the beginning of the dehydration profile measurement of the worn CL matched the expected $M_{EWC}$; in others, it was slightly lower, as reported in the following *Results* section. Thus, measurements began at $M_0$ equal to $M_{EWC}$ or, when it was not possible, from the initially recorded mass $M_0$. As with the unworn CLs, the CLs worn by the subjects had a power of −0.50 D, regardless of the visual correction needs of the wearer. During the wear period, participants either used their glasses or remained unaided, depending on their preference. The average age of the volunteers who wore the CLs for five minutes was 27.1 years (range 24.6–31.2).

The third set of experiments followed the same protocol as the five-minute wear test but extended the wear time to two hours. The average age of the wearers in this group was 25.4 years (range 21.1–33.8).

An overview of the number of samples analyzed in each set of experiments is reported in Table 2.

A parameter depending on time $t$, referred to as valid dehydration ($VD$), was calculated from the measurements, as described by González-Méijome et al. [10] and further developed by Ponzini et al. [11]. $VD$ was defined as:

$$VD = \left[\frac{M_t - M_0}{M_{deh} - M_0}\right] \quad (2)$$

where $M_0$ is the initial mass of the sample in its hydrated state, $M_t$ is the sample mass at time $t$, $M_{deh}$ is the sample mass in the fully dehydrated condition (values reported above in the text). As reported elsewhere [11], the $VD$ profile as a function of time $t$ allows to deduce the water

**Table 1**
List of the investigated CL materials and their parameters as provided by the manufacturers. All investigated CL materials are daily disposable. EWC is the equilibrium water content, $t_c$ is the central thickness of single vision CLs of power equal to −3.00 diopters.

| Material (USAN name) | Etafilcon A | Kalifilcon A |
|---|---|---|
| Brand | 1Day Acuvue Moist | Infuse/Ultra One Day |
| Manufacturer | Johnson&Johnson | Bausch + Lomb |
| FDA classification group (subgroup) | II | V (B) |
| Type of CL | Hydrogel | Silicone-hydrogel |
| EWC (%) | 58 | 55 |
| $t_c$ (mm) | 0.084 | 0.080 |
| Principal monomers | poly-2-hydroxyethylmethacrylate<br>methyl methacrylate<br>polyvinylpyrrolidone | dimethylacrylamide<br>poly-2-hydroxyethylmethacrylate<br>siloxane macromer<br>polyvinylpyrrolidone |

**Table 2**
Number of measured CLs for each experiment (RE: right eye, LE: left eye).

| Sample type | Number of samples | |
|---|---|---|
| | Etafilcon A | Kalifilcon A |
| UNWORN<br>taken from their packaging blister and rinsed in preservative-free saline solution | 11 | 8 |
| WORN FIVE MINUTES<br>taken from their packaging blister and worn for five minutes, measured while hydrated by tears | 8 (RE) + 8 (LE) | 8 (RE) + 8 (LE) |
| WORN TWO HOURS<br>taken from their packaging blister and worn for two hours, measured while hydrated by tears | 14 (RE) + 14 (LE) | 14 (RE) + 14 (LE) |

kinetics coefficient $d$. The time evolution of $VD$ can be modelled using the equation [19]

$$VD = kt^d \quad (3)$$

where $k$ is a constant and $d$ is the water kinetics coefficient describing the water diffusion in the matrix [11]. The coefficient $d$ is the slope of a line if the Eq. (3) is rewritten as

$$Log_{10}(VD) = Log_{10}(k) + dLog_{10}(t) \quad (4)$$

Therefore, $d$ was determined as the slope of the linear regression of the measured values of $Log_{10}(VD)$ with respect to $Log_{10}(t)$, or equivalently, as the first derivative of $Log_{10}(VD)$ with respect to $Log_{10}(t)$.

### 2.3. Statistical analysis

To compare the initial mass $M_0$ between different CL groups and the coefficient $d$ across various groups, the one-tailed Mann-Whitney statistical test was used (SPSS version 2.8; IBM SPSS statistics, USA). The significance statistical threshold was set at 0.05.

## 3. Results

Fig. 1 shows an example of the measured VD as a function of time $t$ on a bi-logarithmic scale. The dashed line represents an unworn CL that was rinsed in saline solution, while the solid line corresponds to a CL worn by a subject for five minutes, measured immediately after removal from the eye.

It is interesting to note that in the initial phase of the measurements, lasting about 90 s (about 1.95 on the logarithmic scale in Fig. 1), the dehydration profiles of worn and unworn samples are quite different. This time interval corresponds to a loss of about 1–2 mg of water, while the curves tend to follow a similar trend in the subsequent phase. This difference between the unworn and worn CL was often observed in the samples investigated. For this reason, considering Eq. (4), the first derivative of $Log_{10}(VD)$ with respect to $Log_{10}(t)$ was calculated in the interval of time corresponding to the first milligram of mass loss for each investigated sample. The average value of the first derivative in this interval represents the mean water kinetics coefficient $d$ in the first phase of water evaporation from the CL.

Table 3 shows the results (average values and standard deviations)

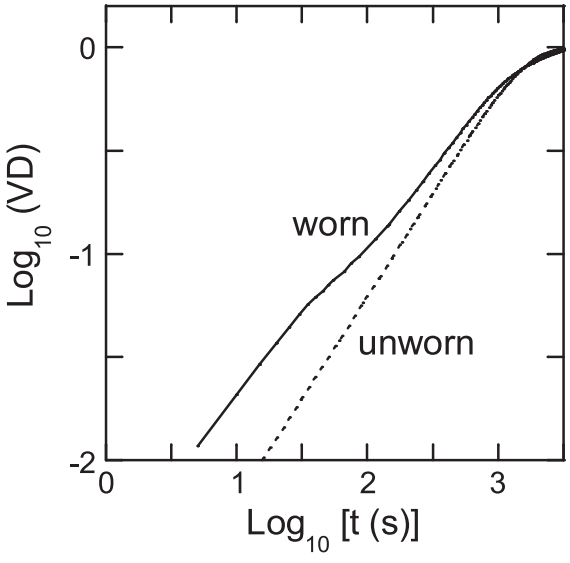


**Fig. 1.** Examples of measured $VD$ curves on either an unworn Kalifilcon A CL rinsed in saline solution (dashed line) or a worn Kalifilcon A CL (solid line) measured immediately after removal from the eye, the worn CL being hydrated by tears.

**Table 3**
Means and standard deviations of the water kinetics coefficient $d$ (Coeff. $d$) corresponding to the first phase of water evaporation (i.e., first milligram of mass loss) and of the initial mass at $t = 0$ ($M_0$), as calculated from the data measured on unworn CLs and on CLs after either five minutes or two hours of wear on the right (RE) or the left eye (LE). In brackets p-values are reported, which were obtained to compare the water kinetics coefficient $d$ between unworn and worn samples and the initial mass between unworn and worn samples. The significance threshold was set at 0.05 (in bold the values below the threshold).

| | | Etafilcon A | | Kalifilcon A | |
|---|---|---|---|---|---|
| | | RE | LE | RE | LE |
| Unworn | *Coeff. d* | 0.97 ± 0.10 | | 1.00 ± 0.01 | |
| | $M_0$ (mg) | 36.61 ± 0.07 | | 28.90 ± 0.09 | |
| Worn five minutes | *Coeff. d* | 0.73 ± 0.09 (**0.0010**) | 0.74 ± 0.05 (**0.0013**) | 0.78 ± 0.12 (**0.0051**) | 0.75 ± 0.04 (**0.0005**) |
| | $M_0$ (mg) | 33.6 ± 1.9 (**0.0007**) | 33.3 ± 1.2 (**0.0002**) | 26.2 ± 1.3 (**0.0027**) | 26.9 ± 1.3 (**0.0005**) |
| Worn two hours | *Coeff. d* | 0.85 ± 0.16 (**0.0375**) | 0.89 ± 0.15 (0.0505) | 0.90 ± 0.12 (**0.0043**) | 0.92 ± 0.12 (0.0808) |
| | $M_0$ (mg) | 36.2 ± 1.0 (0.1038) | 36.2 ± 0.9 (**0.0064**) | 28.3 ± 1.3 (0.4721) | 28.7 ± 0.9 (0.0582) |

found on unworn and worn CLs, separately for right and left eyes, and for the two investigated materials. Both the $d$ coefficient and the initial mass measured at $t = 0$ s are reported in the same table. The same parameters were measured after either 5 min or two hours of wear. The two groups (unworn and worn CLs) were compared in terms of $d$ coefficient and initial mass.

Differences between unworn CLs and CLs worn for two hours can still be observed, although less pronounced compared to the differences between unworn CLs and CLs worn for five minutes.

To quantify the percentage variation, the ratios between the average of $M_0$ and $d$ after 5 min of wear (without distinguishing between right and left eye) and the corresponding averages for unworn CLs were calculated. The same ratios were also calculated for the case of two hours of wear (Fig. 2). It is possible to notice that both $M_0$ and $d$ decrease during the first five minutes of wear (Fig. 2), whereas they go back up in two hours. Interestingly, this is independent from the material.

To better understand the results in Table 3 and Fig. 2, Fig. 3 shows the water kinetics coefficient $d$ vs the mass of each worn sample hydrated by tears. The empty circles (wear of five minutes) fully confirm what was observed regarding the five-minute wear period based on the statistical analysis in Table 3. In fact, it is evident that both the coefficient $d$ and the mass of each sample are lower than the typical values for unworn CLs. On the other hand, the distribution of the full diamonds (wear of two hours) shows that some samples measured after two hours have essentially the same characteristics as the five-minute worn CLs, while other results are consistent with the unworn CLs.

## 4. Discussion

A first point to discuss is the hydration behavior of CLs during wear. The observed reduction in initial water content ($M_0$) after five minutes of wear, compared to unworn CLs, suggests a ~10 % dehydration occurring shortly after lens insertion (Table 3 and Fig. 2). This finding is in agreement with previous measurements of in-vivo CL dehydration [20]. Additionally, the results of this work indicate that CLs worn for two hours tend to have a higher $M_0$ than those worn for only five minutes. While this may seem counterintuitive at first, it aligns with previous findings [21]. For example, in a study using an eye-blink model with an artificial tear solution, Chan et al. found a statistically significant decrease in CL water content after one hour of incubation [21]. However, after 16 h, most CLs regained their initial water content,

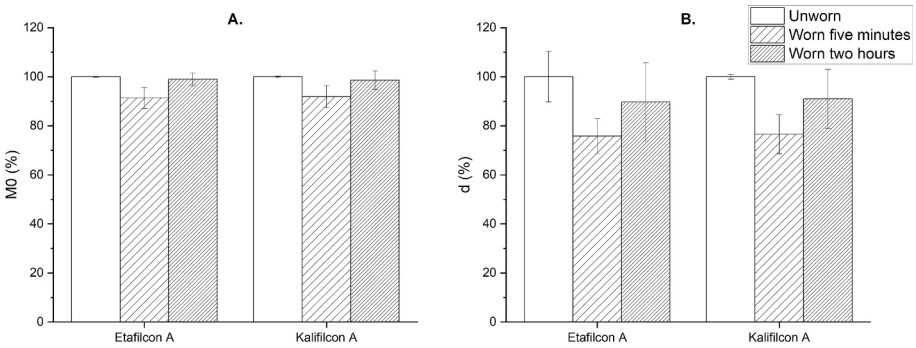


**Fig. 2.** Mean of the initial mass values at $t = 0$ ($M_0$) (panel A) and coefficient $d$ values (panel B) expressed as a percentage of the respective values in unworn CLs. Values were measured on unworn CLs and from CLs after either five minutes or two hours of wear on both the right and the left eye. Error bars represent the standard deviation.

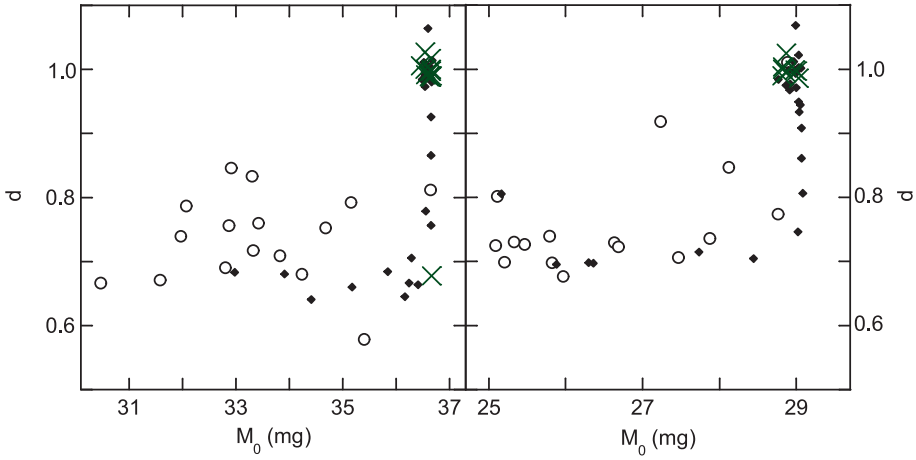


**Fig. 3.** Water kinetics coefficient ($d$) as a function of the initial mass ($M_0$) for unworn Etafilcon A (left panel) and Kalifilcon A CLs (right panel). Data are displayed both for unworn lenses (crosses), and for lenses worn and measured immediately after removal from the eye while still hydrated by tears (empty circles; worn for 5 min; full diamonds: worn for 2 h).

almost matching the original blister pack values. This recovery was absent in a vial-based system [21], suggesting that blinking plays a role in modulating water content changes. Besides blinking, osmotic pressure appears to play a key role [22]. Since tear fluid is more concentrated than the saline storage solution, water initially leaves the CL due to osmotic gradients. As wear continues, the lens begins to reabsorb water from the tear fluid, replenishing some of its lost water content. This absorption reduces the osmotic gradient, leading to a new equilibrium between the lens and the surrounding tear fluid. Additionally, interactions between the tear components and the lens material may further modulate surface evaporation [23–25].

While the dependence of CL hydration on wear duration is consistent with existing literature, the second point of this discussion was less predictable. This work shows that the water kinetics coefficient $d$, measured in vitro on worn CLs, accurately reflects the actual in-vivo CL hydration state during wear. Specifically, CLs exhibiting partial dehydration when removed from the eye (i.e., $M_0 < M_{EWC}$) also exhibit lower $d$ values. This correspondence between the hydration state and the $d$ value opens up the possibility of gaining further insight into the water diffusion mechanisms in the polymer. Unworn CLs display a water kinetics coefficient very close to 1 (Table 3), suggesting non-Fickian dehydration behavior. As discussed in the *Introduction*, this indicates that water evaporation from the outer layer is hindered by rapid polymer shrinkage, whose rate overtakes the diffusion rate of water [13,14]. This behavior requires a flexible polymer network, which is expected in materials in their rubbery state, i.e. above their glass transition temperature [26]. From an in-vivo perspective, this property may be beneficial for the wearer, as it helps prevent excessive dehydration during wear. No statistically significant difference was found between unworn Etafilcon A and unworn Kalifilcon A. This is consistent with findings from a previous study [11], where five unworn materials were compared. In that study, Etafilcon A and Omafilcon A, both hydrogel contact lenses, exhibited similar behaviour, distinct from that of Senofilcon A and Delefilcon A, both silicone-hydrogel contact lenses [11]. Notably, Kalifilcon A, despite being a silicone-hydrogel, displayed a hydration profile reminiscent of the two investigated hydrogel contact lenses, Etafilcon A and Omafilcon A [11].

As shown in Table 3, CLs worn for only five minutes and hydrated with tears exhibit a smaller $d$ value than unworn samples. This suggests a shift toward a more Fickian diffusion behavior, with reduced influence from polymer matrix shrinkage [13,14]. The altered balance between water diffusion and matrix shrinkage may indicate a decrease in polymer flexibility, possibly associated with a transition of the CL outer layer from the rubbery to the glassy state during early in-vivo dehydration as discussed below.

Indeed, several studies support this interpretation. Although Fornasiero et al. focused on different materials (Polymacon, Hilafilcon A, and Balafilcon A), their findings provide valuable insights for interpreting the present results [26]. Their study demonstrated that, at ambient and physiological temperatures, the investigated materials remain in their rubbery state. However, when CLs undergo partial dehydration, a glassy "skin" can form on the outer layer due to a local rubber-to-glass transition. This phenomenon is expected to alter CL properties, including water permeability, on-eye lens movement,

dimensional stability, and oxygen diffusion [26]. Similar findings were reported by Sun et al., who studied poly(2-hydroxyethyl methacrylate) membranes at 37 °C [27]. Their data suggest that glass-rubber transitions complicate water transport in the polymer, likely due to slow polymer network relaxation or reversible immobilization reactions coupled with diffusion [27]. Koffas et al. further explored the surface mechanical properties of poly(hydroxyethyl methacrylate) using atomic force microscopy [28]. Their results show that at relative humidity levels below 60 %, the lens surface loses water and behaves like a glassy polymer. In contrast, higher humidity levels result in reduced surface evaporation and increased softening, as water plasticizes the hydrogel surface [28]. Together, these studies suggest a plausible scenario for the trends observed in the present work. In-vivo dehydration may lead to a localized reversible rubber-to-glass transition at the lens surface, resulting in a rigid outer layer (a glassy "skin") that reduces matrix shrinkage. As a result, water is able to diffuse more freely within the polymeric matrix, leading to lower *d* values compared to unworn CLs. This shift may serve as an early indicator of non-optimal hydration and could potentially impact lens performance and wearer comfort. Importantly, these findings confirm that the water kinetics coefficient *d* is not only a meaningful descriptor of water diffusion dynamics but also a sensitive marker of hydration state under near-physiological conditions.

As already discussed for unworn CLs, also for worn CLs no differences emerged between the two materials. The comparison of the *d* coefficients for the two materials was conducted separately for the right and left eye, and separately for the five-minute wear and the two-hour wear. All four calculated p-values were well above 0.05.

Finally, the subject, not the eye, was the statistical unit of analysis. Because the two eyes of the same subject are not independent, analyses were conducted separately for right and left eyes but never combined in a way that would inflate the sample size (Table 3). A preliminary analysis showed that, for all comparisons between right and left eyes, both after five minutes and after two hours of lens wear, the measurements were strongly correlated (Spearman's rho $>0.85$, $p < 0.0001$). In such cases, reporting a single eye, chosen at random, would be sufficient. In the present study, however, the statistical analyses from both eyes in parallel are reported as well, not to increase statistical power, but to provide transparent evidence of the internal consistency of the results and to show that the conclusions remain unchanged regardless of which eye is considered.

## 5. Conclusions

This study investigates the dehydration kinetics of Etafilcon A and Kalifilcon A CLs under controlled in-vitro conditions, comparing unworn CLs hydrated in saline solution and CLs worn and hydrated by natural tears.

The findings underscore that the degree of lens dehydration and the associated changes in physical parameters vary substantially depending on how long the lens has been worn. Notably, a measurable decrease in water content occurs within the first five minutes of wear. However, after a longer period of wear (two hours), approximately half of the analyzed CLs exhibited rehydration, suggesting the establishment of a new hydration equilibrium. The finding that only a portion of the contact lenses exhibited rehydration after two hours of wear raises important questions. The aim of this preliminary study was to explore early dehydration behavior rather than investigate inter-individual variability in ocular surface parameters. As such, no stratification or subgroup analysis based on sex, age, tear film quality, or ocular surface condition was performed. All participants were screened to ensure they had no known ocular surface disease or tear film abnormalities. The observed variability in rehydration could be influenced by individual factors, such as tear film composition, blink rate, or lens-lid interaction. The absence of rehydration in some lenses may indicate a persistent imbalance in tear-lens fluid exchange or material-dependent differences in water retention under specific ocular conditions. These hypotheses merit further investigation and could represent a direction for future studies, which should include a larger, more diverse cohort and consider subject-specific ocular parameters to better understand these divergent hydration patterns.

The in-vitro water kinetics coefficient *d* emerged as a reliable and sensitive indicator of the actual on-eye hydration state of the CLs, providing a meaningful tool for understanding the underlying mechanisms of water diffusion. According to a model that accounts for water diffusion within the polymer matrix and the polymer reaction, lower *d* values indicate a reduced ability of the polymer to undergo shrinkage in order to limit evaporation. In agreement with this model, lower *d* values were here found for on-eye partially dehydrated CLs. This modification of the material properties that promote free water diffusion is consistent with a possible rubber-to-glass transition in the outer lens layer. Such a transition would reduce the polymer flexibility and its ability to shrink, thereby decreasing its capacity to retain water.

This study provides preliminary insights into the early dehydration kinetics of soft contact lenses during wear, which are expected to contribute to a deeper understanding of water transport and dehydration behavior in CLs during wear. A key limitation is the use of different subjects for the 5-minute and 2-hour wear groups, which may introduce variability related to tear film composition. However, efforts were made to reduce bias by selecting healthy participants and standardizing environmental and handling conditions. As a next step, future research should adopt a longitudinal design to monitor dehydration profiles within the same individuals over extended wear periods, from the initial minutes up to 8 h. This approach would enable a more precise understanding of how material properties evolve in relation to tear film dynamics under real-world conditions.

Further research is needed to explore novel strategies for maintaining surface hydration and biocompatibility in CLs. Potential approaches may include preferential segregation or grafting of hydrophilic components to the surface. A similar in-vitro study on other types of worn CLs could help evaluate the effectiveness of current or future strategies aimed at mitigating surface dehydration. Future studies should also investigate the role of adsorbed proteins on both surfaces of a CL, as protein layers may modulate water uptake and evaporation, potentially increasing the near-surface water content by reducing water transport across the lens.

## Declaration of competing interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

## References

[1] Brennan NA, Lowe R, Efron N, Harris MG. In vivo dehydration of disposable (Acuvue) contact lenses. Optom Vis Sci 1990;67:201–3. https://doi.org/10.1097/00006324-199003000-00010.

[2] Fonn D, Situ P, Simpson T. Hydrogel lens dehydration and subjective comfort and dryness ratings in symptomatic and asymptomatic contact lens wearers. Optom Vis Sci 1999;76:700–4. https://doi.org/10.1097/00006324-199910000-00021.

[3] Efron N. Contact lens wear is intrinsically inflammatory. Clin Exp Optom 2017;100:3–19. https://doi.org/10.1111/cxo.12487.

[4] Recchioni A, Mocciardini E, Ponzini E, Tavazzi S. Viscoelastic properties of the human tear film. Exp Eye Res 2022;219:109083. https://doi.org/10.1016/j.exer.2022.109083.

[5] Tavazzi S, Tonveronachi M, Fagnola M, Cozza F, Ferraro L, Borghesi A, et al. Wear effects on microscopic morphology and hyaluronan uptake in siloxane-hydrogel contact lenses. J Biomed Mater Res B Appl Biomater 2015;103:1092–8. https://doi.org/10.1002/jbm.b.33278.

[6] Bettuelli M, Trabattoni S, Fagnola M, Tavazzi S, Introzzi L, Farris S. Surface properties and wear performances of siloxane-hydrogel contact lenses. J Biomed Mater Res B Appl Biomater 2013;101:1585–93. https://doi.org/10.1002/jbm.b.32901.

[7] Jones L, Efron N, Bandamwar K, Barnett M, Jacobs DS, Jalbert I, et al. TFOS lifestyle: impact of contact lenses on the ocular surface. Ocul Surf 2023;29:175–219. https://doi.org/10.1016/j.jtos.2023.04.010.

[8] Tranoudis I, Efron N. Water properties of soft contact lens materials. Contact Lens Anterior Eye 2004;27:193–208. https://doi.org/10.1016/j.clae.2004.08.003.
[9] Nichols JJ, Willcox MDP, Bron AJ, Belmonte C, Ciolino JB, Craig JP, et al. The TFOS international workshop on contact lens discomfort: executive summary. Invest Ophthalmol Vis Sci 2013;54:7–13. https://doi.org/10.1167/iovs.13-13212.
[10] González-Méijome JM, López-Alemany A, Almeida JB, Parafita MA, Refojo MF. Qualitative and quantitative characterization of the in vitro dehydration process of hydrogel contact lenses. J Biomed Mater Res B Appl Biomater 2007;83B:512–26. https://doi.org/10.1002/jbm.b.30824.
[11] Ponzini E, Maspero F, Galli A, Tavazzi S. In-vitro dehydration kinetics coefficient of Kalifilcon a and other contact lens materials. Sci Rep 2024;14:7870. https://doi.org/10.1038/s41598-024-55937-2.
[12] Ponzini E, Borghesi A, Galli A, Maspero F, Rizzo GC, Trabattoni S, et al. Dehydration and surface friction of Kalifilcon a and Senofilcon a contact lenses. Nuovo Cimento Della Soc Ital Fis C 2024;47. https://doi.org/10.1393/ncc/i2024-24334-2.
[13] Frisch HL. Sorption and transport in glassy polymers–a review. Polym Eng Sci 1980;20:2–13. https://doi.org/10.1002/pen.760200103.
[14] Yang C, Xing X, Li Z, Zhang S. A comprehensive review on water diffusion in polymers focusing on the polymer–metal interface combination. Polymers 2020;12:138. https://doi.org/10.3390/polym12010138.
[15] Pritchard N, Fonn D. Dehydration, lens movement and dryness ratings of hydrogel contact lenses. Ophthalmic Physiol Opt 1995;15:281–6. https://doi.org/10.1016/0275-5408(95)00004-W.
[16] Refojo MF. Tear evaporation considerations and contact lens wear. Consid. Contact Lens Use Adverse Cond. Proc. Symp., National Academies Press (US); 1991.
[17] Insua Pereira E, Lira M. Comfort, ocular dryness, and equilibrium water content changes of daily disposable contact lenses. Eye Contact Lens 2018;44(Suppl 2):S233–40. https://doi.org/10.1097/ICL.0000000000000441.
[18] González-Méijome JM, da Silva AC, Neves H, Lopes-Ferreira D, Queirós A, Jorge J. Clinical performance and “ex vivo” dehydration of silicone hydrogel contact lenses with two new multipurpose solutions. Contact Lens Anterior Eye 2013;36:86–92. https://doi.org/10.1016/j.clae.2012.10.071.
[19] Ritger PL, Peppas NA. A simple equation for description of solute release I. Fickian and non-fickian release from non-swellable devices in the form of slabs, spheres, cylinders or discs. J Control Release 1987;5:23–36. https://doi.org/10.1016/0168-3659(87)90034-4.
[20] Krysztofiak K, Szyczewski A. Study of dehydration and water states in new and worn soft contact lens materials. Opt Appl 22014 ISSN 1429-7507 2014. https://doi.org/10.5277/OA140206.
[21] Chan VWY, Phan C-M, Walther H, Ngo W, Jones L. Effects of temperature and blinking on contact lens dehydration of contemporary soft lens materials using an in vitro blink model. Transl Vis Sci Technol 2021;10:11. https://doi.org/10.1167/tvst.10.8.11.
[22] Fatt I. A predictive model for dehydration of a hydrogel contact lens in the eye. J Br Contact Lens Assoc 1989;12:15–31. https://doi.org/10.1016/S0141-7037(89)80017-5.
[23] Vishnubhatla S, Borchman D, Foulks GN. Contact lenses and the rate of evaporation measured in vitro; the influence of wear, squalene and wax. Contact Lens Anterior Eye 2012;35:277–81. https://doi.org/10.1016/j.clae.2012.07.008.
[24] Ponzini E, Tavazzi S, Musile G, Tagliaro F, Grandori R, Santambrogio C. Contact lens wear induces alterations of lactoferrin functionality in human tears. Pharmaceutics 2022;14:2188. https://doi.org/10.3390/pharmaceutics14102188.
[25] Ponzini E, Scotti L, Zambon A, Santambrogio C, Grandori R, Borghesi A, et al. Lactoferrin as a biomarker of ocular diseases and contact lens discomfort. Nuovo Cimento Della Soc Ital Fis C 2023;46. https://doi.org/10.1393/ncc/i2023-23173-y.
[26] Fornasiero F, Ung M, Radke CJ, Prausnitz JM. Glass-transition temperatures for soft-contact-lens materials. Dependence on water content. Polymer 2005;46:4845–52. https://doi.org/10.1016/j.polymer.2005.03.084.
[27] Sun Y-M. Sorption/desorption properties of water vapour in poly(2-hydroxyethyl methacrylate): 2. Polymer: Two-stage sorption models; 1996.
[28] Koffas TS, Opdahl A, Marmo C, Somorjai GA. Effect of equilibrium bulk water content on the humidity-dependent surface mechanical properties of hydrophilic contact lenses studied by atomic force microscopy. Langmuir 2003;19:3453–60. https://doi.org/10.1021/la026719x.