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Performance Analysis of Medical Syringe and Infusion Pumps: (Factorial Experiments Case Study on the Combined Effects of Syringe Size, Infusion Line Length and Tubing Compliance)

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Abstract

Medication errors are among the leading causes of serious adverse events in hospitalized patients, and a considerable number of deaths each year can be traced back to them. In syringe and infusion pumps, delivery accuracy is a safety-critical property, especially in intensive care and neonatal environments. Importantly, the accuracy of these devices is not dictated by the motor and drive mechanism alone. A group of external factors, namely the size and mechanical compliance of the mounted syringe, the length and internal diameter of the infusion line, and the material behavior of the tubing, exert a strong and frequently underestimated influence on performance.

This study sets out an integrated experimental and statistical investigation of the variables that shape syringe pump performance. Within a full factorial design, several analytical techniques were applied. The data set comprised more than 300 observations spanning three syringe sizes, three programmed flow rates, several tubing types, and two infusion line lengths. The results yield evidence-based guidance for nurses and clinical engineers. They confirm that syringe size, tubing compliance, and line length are each statistically significant determinants of pump performance, with the largest effects appearing at the lowest flow rate (1.0 mL/h). Taken together, the findings show that larger syringe diameters magnify the effect of mechanical play in the drive train; that long infusion lines add both fluidic resistance and volumetric compliance, the latter being the dominant cause of delayed flow onset at low rates; and that compliant tubing stores elastic strain energy during start-up, producing a delayed onset of flow together with transient under-delivery.

Keywords Medical Devices; Syringe and Infusion Pumps; Flow Rate Accuracy; Performance Analysis.

1. Introduction

Nowadays, medical devices are widely used to improve health outcomes and healthcare delivery. Even the best of these devices is affected by wear and tear, environ-

mental changes, and other external factors, which can reduce their accuracy over time. The accuracy of instrument readings, one of the most important parameters for medical devices, indicates how closely the equipment's



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outcome matches the true value. [1]

Continuous intravenous drug infusion is one of the most widely used treatment modalities in modern medicine, and accurate, sustained delivery of a drug to the patient over an extended period is therefore essential. Syringe and infusion pumps are precision electromechanical devices engineered to administer drugs at accurate, programmable flow rates. They are indispensable wherever precise and continuous delivery is required, such as critical care units, neonatal intensive care units (NICUs), operating theatres, and outpatient chemotherapy suites. [1, 2]

Certain clinical situations push the fluidic system toward its limits. When caring for a patient with a highly infectious and potentially fatal pathogen, staff may site the pump outside the patient's room and connect it through a longer extension line in order to limit contact and reduce the risk of transmission. In other cases, a larger syringe is selected to cut down the frequency of syringe changes. In both scenarios the compliance of the system rises, and the effect becomes most pronounced at low flow rates. [2] The international standard IEC 60601-2-24 sets out baseline performance requirements for infusion pumps, covering flow rate accuracy, bolus delivery, and occlusion detection. The standard, however, does not require testing across every syringe brand, every tubing length, or every compressibility scenario that may be met at the bedside. [3, 4]

The fluidic system, comprising the syringe, the infusion line, and any ancillary components, couples with the pump drive mechanism and, taken as a whole, behaves as a dynamic mechanical system. These elements interact in complex and non-linear ways, particularly at low flow rates. A clear understanding of this interaction is central to optimizing pump configuration, reducing medication errors, and improving patient safety. In the present in vitro study, we examine these issues in detail, with particular attention to how different syringe sizes and infusion line lengths affect the behavior of the infusion system. [5-9]

2. Materials and Methods

2.1 Study Design

An in vitro full factorial experimental design was adopted to characterize how variations in infusion line length and syringe size affect the compliance and overall performance of the infusion system.

Four independent variables were manipulated: (1) syringe size, drawn from the BD Plastipak range (10, 20, and 50 mL); (2) programmed flow rate (1.0, 5.0, and 10.0 mL/h); (3) tubing type (PVC standard bore, PVC microbore, PU standard bore, PU microbore, and a silicone reference); and (4) infusion line length (75 cm and 150 cm).

Within the test rig, the infusion tubing and syringes were treated as interchangeable components, since they are the elements that govern system compliance. Start-up delay was operationally defined as the interval from pump activation to the first detection of a continuous increase in mass at a rate corresponding to at least 90% of the programmed flow.

All trials were carried out with a JMS SP-500X syringe pump using a saline-filled syringe at a controlled temperature of 22 ± 0.5 °C. Before data collection began, the pump was calibrated against an infusion analyzer (IDA-4, Fluke Biomedical). Each variable combination was repeated multiple times: 20 replicates per cell at 1 mL/h for the accuracy measurements, and 15 replicates per cell for the start-up delay measurements.

Flow rate accuracy was determined gravimetrically using a Mettler-Toledo analytical balance with a resolution of 0.1 mg. Gravimetric weighing is generally regarded as the benchmark approach for quantifying both flow rate accuracy and start-up delay. Any visible air bubbles were eliminated beforehand by gentle tapping and flushing of the line.

2.2 Statistical Analysis

Descriptive statistics were computed for every outcome variable. A two-way ANOVA was used to evaluate the main effects of syringe size and flow rate, together with their interaction, on the percentage flow error. Pearson correlation coefficients were calculated across all continuous predictors and outcomes, and a multiple linear regression model was then fitted with percentage flow error as the dependent variable. Before the parametric analyses, the normality of the within-cell residuals was examined with the Shapiro–Wilk test and the homogeneity of variances across cells with Levene's test; because the coefficient of variation ranged from 0.58% to 9.80% between cells, the outcome of the variance-homogeneity check is reported together with the ANOVA, and a Welch-corrected analysis was used as a sensitivity check where the assumption was not met. Significant main effects were followed by Tukey's HSD post-hoc comparisons, and effect estimates are reported with 95% confidence intervals alongside the corresponding p-values. All analyses were performed in R, version 4.1.3 (2022).

Percentage flow rate error was expressed relative to the programmed rate:

$$\text{Error (\%)} = [(Q_{\text{delivered}} - Q_{\text{programmed}}) / Q_{\text{programmed}}] \times 100$$

For each experimental cell the sample mean and standard deviation were obtained, and the coefficient of variation (CV) derived as follows:

$$\text{CV (\%)} = (\text{SD} / \text{mean}) \times 100$$

Table 1. Flow rate accuracy by syringe size and programmed flow rate (line: 150 cm PVC standard bore, n = 20 per condition).

Syringe (mL)	Flow Rate (mL/h)	Mean Delivered (mL/h)	SD (mL/h)	Rel. Error (%)	CV (%)
10	1.0	0.962	0.031	−3.80	3.22
10	5.0	4.991	0.044	−0.18	0.88
10	10.0	10.023	0.058	+0.23	0.58
20	1.0	0.941	0.052	−5.90	5.53
20	5.0	4.978	0.063	−0.44	1.27
20	10.0	10.031	0.071	+0.31	0.71
50	1.0	0.908	0.089	−9.20	9.80
50	5.0	4.963	0.082	−0.74	1.65
50	10.0	10.047	0.094	+0.47	0.94

3. Results

3.1 Effect of Syringe Size and Compliance

3.1.1 Mechanical Geometry

A syringe pump advances the plunger at a tightly controlled linear velocity, which is converted into a volumetric flow rate through the cross-sectional area of the syringe barrel. Because that area scales with the square of the inner radius, a larger syringe delivers proportionally more volume for the same plunger travel. Mechanical play in the drive train, usually termed backlash, is the small gap between the lead screw and the drive nut (or between other coupled parts) that must be closed before forward motion is actually passed to the plunger.

In a 10 mL syringe with an inner diameter of roughly 15.9 mm, the barrel cross-section is about 199 mm², so 0.1 mm of backlash translates into a volume error of about 20 µL. The same 0.1 mm of backlash in a 50 mL syringe, whose inner diameter is approximately 28.6 mm and whose cross-section is about 642 mm², produces a volume error more than three times as large, on the order of 64 µL. At low programmed flow rates these volume discrepancies manifest as a noticeable temporal delay in the onset of drug delivery.

3.1.2 Syringe Barrel Compliance

Syringe barrels deform elastically under internal pressure. During the initial phase, part of the applied force is taken up by the elastic barrel wall and the plunger seal rather than being converted directly into fluid flow. The

magnitude of this compliance effect differs between manufacturers because of variations in wall thickness, material grade, and the design of the plunger gasket.

System compliance can postpone the effective onset of flow by tens of seconds at low flow rates, and the delay grows as syringe size increases. Where a pump incorporates a compensation algorithm calibrated for one specific syringe geometry and compliance profile, the practice of swapping syringe brands on a given pump model becomes a clinically relevant concern.

3.2 Infusion Line Characteristics

The infusion line itself behaves as a fluidic resistor. According to the Hagen-Poiseuille relationship for laminar flow, the pressure drop along the line is proportional to its length, to fluid viscosity, and to the volumetric flow rate, while being inversely proportional to the fourth power of the inner radius. As a consequence, even a small reduction in inner diameter produces a large rise in flow resistance.

In practical terms, longer infusion lines generate greater fluidic resistance. At low flow rates the steady-state pressure drop produced by this resistance remains small, because the pump drive develops more than enough pressure to overcome it; the delayed onset of delivery observed at low rates is therefore governed by the additional volumetric compliance of the longer line rather than by its resistance. At higher flow rates, or when viscous infusates such as blood products or lipid emulsions are

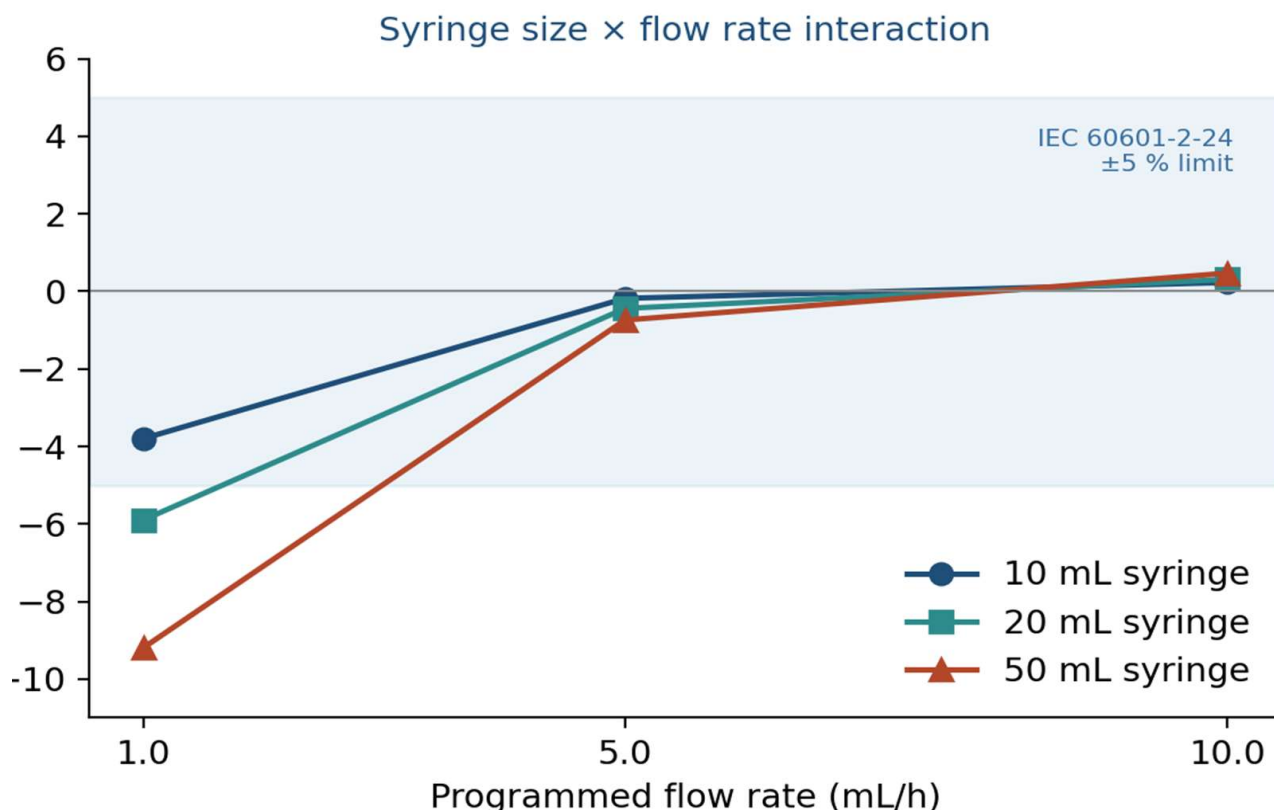


Fig 1. Figure 1. Interaction between syringe size and programmed flow rate for relative flow rate error (150 cm PVC standard-bore line, $n = 20$ per cell). The shaded band marks the $\pm 5\%$ accuracy limit of IEC 60601-2-24.

used, the resistance of a long line may exceed the pump's maximum delivery pressure, triggering a false-positive occlusion alarm or an unintended reduction in flow.

Tubing diameter directly affects priming volume, hydraulic resistance, and the volumetric compliance of the line. Manufacturers use several standard inner-diameter configurations, ranging from roughly 2.0 mm for standard sets down to about 0.5 mm for the microbore tubing used in pediatric applications. Standard tubing with a larger inner diameter offers lower resistance but a larger internal volume, so a greater fluid volume is needed before steady-state flow is reached at the catheter tip. Microbore tubing reduces dead volume, an advantage for dose accuracy, but its higher resistance can introduce a substantial pressure drop at the distal end.

3.3 Measurement Analysis

Delivery proceeds through three phases. During the start-up phase the pump under-delivers as syringe and tubing compliance are charged, so the patient receives less drug than intended. A brief overshoot may follow in the transition phase as the stored mechanical energy is released. In the final, steady-state phase the pump reaches its pro-

grammed rate within the accuracy specification.

3.3.1 Flow Rate Accuracy Analysis

IEC 60601-2-24 specifies that infusion pumps should hold flow rate accuracy within $\pm 5\%$ of the programmed value at steady state. Table 1 reports the flow rate accuracy data for all combinations of syringe size and flow rate at a line length of 150 cm using PVC standard-bore tubing, and Figure 1 displays the same data as an interaction plot.

At 1.0 mL/h, the magnitude of the relative error rose sharply with syringe size: -3.80% for the 10 mL syringe, -5.90% for the 20 mL, and -9.20% for the 50 mL, the last two falling outside the $\pm 5\%$ limit of IEC 60601-2-24. At 10.0 mL/h, by contrast, all three syringe sizes performed within $\pm 0.5\%$ of target.

3.3.2 Start-up Delay Analysis

Table 2 presents start-up delay measurements across syringe sizes, tubing types, and line lengths, and Figure 2 summarises the low-flow condition graphically. The most concerning result was a delay of 112.7 ± 7.2 s, recorded with a 50 mL syringe, PVC standard-bore tubing, and a

150 cm line at 1.0 mL/h. Switching to polyurethane (PU) microbore tubing under otherwise identical conditions cut this delay to 31.4 ± 4.1 s, a reduction of roughly 72.1%. The shortest delays across all configurations occurred at 5.0 mL/h with PU microbore tubing, where the mean delay was 8.1 s.

4. Statistical Analysis

4.1 Two-Way ANOVA (Syringe Size $\times$ Flow Rate)

Table 3 summarises the two-way ANOVA used to test the effects of syringe size and programmed flow rate on percentage flow error. Both main effects were statistically significant: syringe size ($F(2,160) = 48.72$, $p < 0.001$,

Table 1. Start-up delay by syringe size, tubing type (PVC standard, PU microbore), and line length ($n = 15$ per condition).

Syringe (mL)	Flow Rate (mL/h)	Tubing Type	Line Length (cm)	Mean Delay (s)	SD (s)
10	1.0	PVC Standard	75	22.4	3.7
10	1.0	PVC Standard	150	38.6	5.1
20	1.0	PVC Standard	75	41.3	4.8
20	1.0	PVC Standard	150	67.8	6.3
50	1.0	PVC Standard	75	74.2	5.9
50	1.0	PVC Standard	150	112.7	7.2
50	1.0	PU Microbore	150	31.4	4.1
50	5.0	PVC Standard	150	18.3	2.8
50	5.0	PU Microbore	150	8.1	1.6

Table 3. Two-way ANOVA: effect of syringe size and flow rate on flow rate error (%).

Source of Variation	SS	df	MS	F	p-value	η^2
Syringe Size	14.831	2	7.416	48.72	< 0.001	0.524
Flow Rate	22.647	2	11.324	74.38	< 0.001	0.601
Syringe $\times$ Flow Rate	3.214	4	0.804	5.28	0.0011	0.113
Residual (Error)	24.378	160	0.152	—	—	—
Total	64.070	168	—	—	—	—

sum of squares; df = degrees of freedom; MS = mean square; η^2 = partial eta-squared (effect size); $\alpha = 0.05$.

Table 4. Pearson correlation matrix of continuous study variables.

Flow Error (%)	1.000	0.847**	0.791**	0.613**	0.582**
Start-up Delay (s)	0.847**	1.000	0.903**	0.774**	0.691**
Syringe Size (mL)	0.791**	0.903**	1.000	0.421*	0.398*
Line Length (cm)	0.613**	0.774**	0.421*	1.000	0.527**
Tubing Compliance (%)	0.582**	0.691**	0.398*	0.527**	1.000

* $p < 0.05$; ** $p < 0.001$ (two-tailed). Values are Pearson r coefficients.

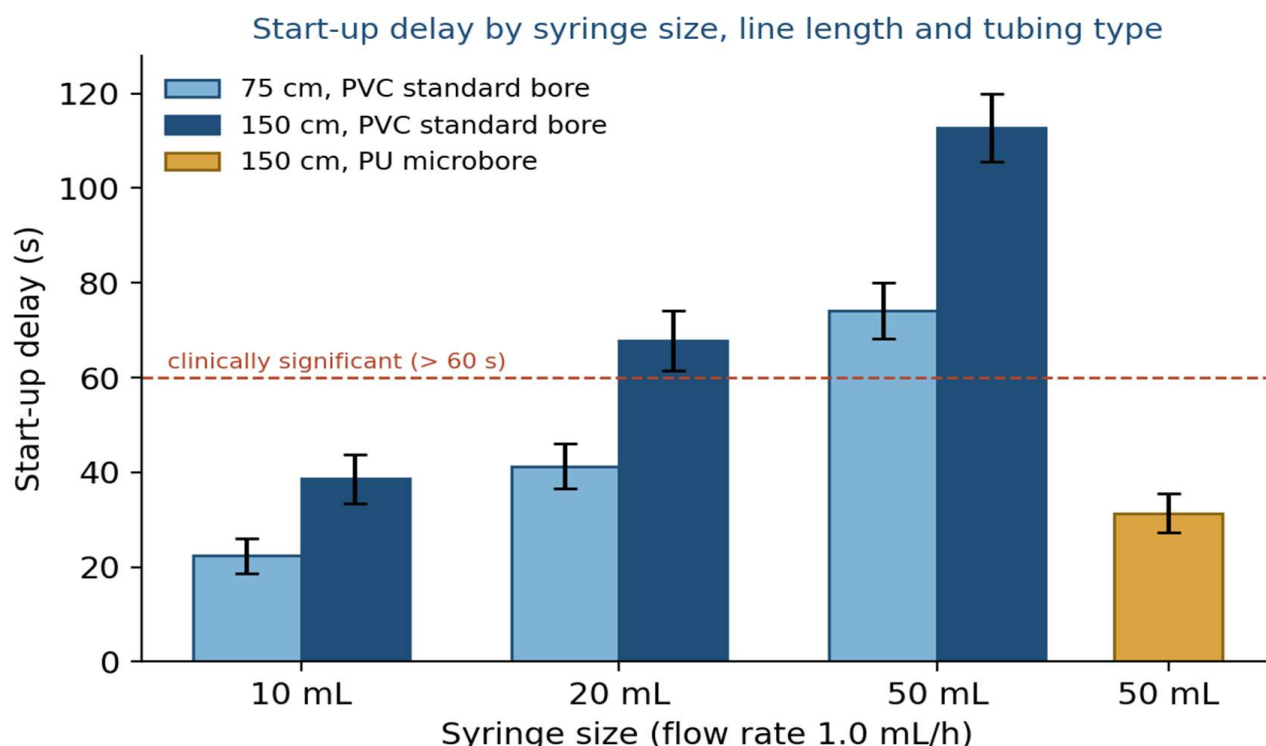


Fig. 2 Mean start-up delay ($\pm$ SD, $n = 15$ per cell) at 1.0 mL/h as a function of syringe size, line length and tubing material. The dashed line marks the 60 s threshold adopted here as clinically significant.

Table 5. Summary of key variables affecting syringe pump performance.

Variable	Primary Effect	Clinical Impact
Large syringe size	Amplified backlash error	Start-up delay, dose inaccuracy
Syringe compliance	Energy absorption at start-up	Flow-onset latency at low rates
Long infusion line	Increased fluidic resistance	Occlusion alarms, flow reduction
Large tubing diameter	High dead volume; slow equilibration	Delayed concentration change
PVC (compliant) tubing	Volumetric compliance, tube expansion	Start-up delay

$\eta^2 = 0.524$) and flow rate ($F(2,160) = 74.38$, $p < 0.001$, $\eta^2 = 0.601$). The interaction term was likewise significant ($F(4,160) = 5.28$, $p = 0.0011$, $\eta^2 = 0.113$), indicating that the influence of syringe size on flow error becomes more pronounced as flow rate falls, most notably at 1.0 mL/h.

4.2 Pearson Correlation Matrix

Pearson correlation analysis was run across all continuous experimental variables, with results given in Table 4. Start-up delay correlated most strongly with syringe size ($r = 0.903$), followed by line length ($r = 0.774$) and tubing compliance ($r = 0.691$). Flow error was most closely associated with start-up delay ($r = 0.847$) and syringe size ($r = 0.791$). The inter-correlations among predictors remained

within an acceptable range for regression modelling, indicating no problematic multicollinearity.

Each entry in the matrix was obtained from the standard product-moment formula:

$$r = \frac{\sum[(x_i - \bar{x})(y_i - \bar{y})]}{\sqrt{[\sum(x_i - \bar{x})^2 \cdot \sum(y_i - \bar{y})^2]}}$$

5. Discussion

The findings of this study provide a robust quantitative foundation for understanding the multifactorial nature of variability in syringe pump performance. The observation that 50 mL syringes operating at 1.0 mL/h overshot the $\pm 5\%$ accuracy limit by almost four percentage points is clinically alarming. Syringe size and flow rate emerged as the two dominant predictors of error.

The strong correlation between syringe size and start-up delay ($r=0.903$) confirms that compliance-driven energy absorption scales with the cross-sectional area of the syringe barrel. Notably, replacing PVC standard-bore tubing with PU microbore tubing reduced start-up delay by more than 72% under matched conditions, demonstrating that tubing choice is a modifiable, high-impact lever in clinical pump configuration. The significant interaction between syringe size and line length ($p=0.021$) reinforces the point that these variables should not be optimized in isolation.

Table 5 summarizes the key variables that influence syringe pump performance, together with their consequences.

Drawing on the analysis above, the following graded, evidence-informed recommendations are proposed:

- Strong: For low flow rate infusions at ≤ 2 mL/h, use the smallest clinically appropriate syringe. The 10 mL syringe produced 58.7% less flow error than the 50 mL syringe at 1.0 mL/h ($p < 0.001$).
- Strong: Standardise syringe brands within each clinical unit and validate them against the pump model in use, since cross-brand compliance variability propagates directly into delivery error.
- Strong: For low-flow critical care infusions, prefer PU microbore tubing over PVC standard bore. Under otherwise identical conditions (50 mL syringe, 150 cm line, 1.0 mL/h) this change reduced start-up delay by 72.1%, from 112.7 ± 7.2 s to 31.4 ± 4.1 s.
- Moderate: Keep infusion lines as short as practicable by positioning pumps close to the IV access point. Each additional 75 cm of PVC standard-bore tubing adds between 16.2 s (10 mL syringe) and 38.5 s (50 mL syringe) to start-up delay, a mean of 27.1 s across the three syringe sizes, the increase itself scaling with syringe size ($p < 0.001$). Where long lines cannot be avoided, microbore tubing should be the default choice.

5.1 Limitations

Several limitations should be borne in mind when interpreting these results. All measurements were obtained with a single pump unit of one model (JMS SP-500X) and a single syringe brand (BD Plastipak), so the absolute magnitudes reported here should not be transferred to other pump–syringe combinations without local verification. The gravimetric measurements were not protected by an evaporation trap or a surface oil layer, which may introduce a small negative bias at the lowest flow rate, and no independent repeatability study of the balance was conducted beyond the initial calibration against the infusion analyser. Testing was carried out in vitro with saline

at a fixed temperature of 22 ± 0.5 °C and against atmospheric back-pressure, whereas clinical infusions involve viscous drug solutions, variable venous back-pressure, vertical displacement between pump and access site, and co-infusion through shared lumens, all of which may amplify the effects described.

6. Conclusion

The performance of syringe pumps and infusion pumps in clinical use is a multi-variable problem that extends well beyond the intrinsic accuracy of the pump drive mechanism. This study provides a systematic quantitative characterization of the combined effects of syringe size, tubing type, line length, and flow rate on syringe pump delivery performance. These variables interact in ways that can substantially degrade drug delivery accuracy, particularly at low flow rates and during rate transitions. The strongest predictor of error at clinically critical low flow rates is syringe size, followed by tubing compliance and line length. Key findings confirm that long infusion lines increase both fluidic resistance and volumetric compliance and thereby delay the onset of delivery, that larger syringe diameters amplify the effect of mechanical play in the drive mechanism, and that compliant tubing absorbs elastic strain energy during pump start-up, resulting in delayed flow onset and transient under-delivery.

Future work should focus on the development of standardized and clinically representative pump performance test configurations, with incorporation of real-time pressure monitoring into pump control algorithms, and the exploration of active compliance compensation techniques that could reduce start-up delay without requiring hardware modifications to the infusion line.

Funding

This research received no specific grant from any funding agency in the public, commercial, or not-for-profit sectors.

Declarations

Ethics approval and consent to participate

Not applicable.

Consent for publication

Not applicable.

Competing interests

The authors declare that they have no competing interests.

Received: Jun. 2026 Accepted: Sep. 2026

Published online: Sep. 2026

DOI: 10.22034/ASAS.2026.588507.1113

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